Biomechanics of a Posterior Substituting Knee During a Simulated Squatting Motion

Undergraduate Honors Thesis

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by

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Abstract

Total knee arthroplasty (TKA) is a common treatment for osteoarthritis (OA). However, there is a large range of postoperative outcomes for TKA patients. Many of these patients find it difficult to perform daily activities such as walking up stairs. Studies have found that error in surgical technique is the most common reason for a revision TKA and may contribute to these undesirable postoperative outcomes. Therefore, establishing the parameters for "optimal" prosthetic component alignment and positioning can help orthopaedic surgeons to ensure a highly functional postoperative outcome for their patients. An important first step in establishing the "optimal" range of alignments is to first determine the effects of malalignment of the femoral and tibial components on knee biomechanics. My research serves as this first step by determining how the alignment of a posterior substituting knee (PS) impacts the biomechanics of the knee during a simulated squatting motion in the Oxford Rig. Over 200 simulations were performed varying the alignment of the femoral and tibial components in 6° varus to 6° valgus, 20° internal to 20° external, and 5° anterior slope to 10° posterior slope. A regression analysis was then completed to determine the relationship between component alignment and outcome variables. The variables that were focused on include MCL, LCL and knee kinematics at 20^{o} and 90^{o} of flexion angle and quadriceps muscle force at 120^{o}. Component alignments with a $p \leq 0.05$ and coefficients greater than an absolute value of 1 were considered to be significant. From this, it was found that alignment of the tibial component in the frontal plane and the femoral component in both the transverse and frontal planes have the most significant effect on functional outcomes. Further, an equation was found to create an estimated curve for the kinematics. This research suggests that in order to improve functional postoperative outcomes for their patients, surgeons should focus on achieving appropriate femoral alignment in both the transverse and frontal planes and tibial alignment in the frontal plane.

Acknowledgements

There are many people I would like to thank for their help on this project. Without their support, this project could not have been a success. First and foremost, I would like to thank Dr. Robert Siston. When I walked into his office last year, my life changed forever. In this meeting, I not only discovered my future career, but I discovered my passion. In the past year, I have learned more than I could have imagined. Words cannot describe how grateful I am for his constant support and guidance. He pushes me hard and challenges me, and I have grown because of it. None of this would have been possible without him. I am so grateful to have him as my advisor and I am so excited to continue my work with him next year.

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Chapter 1: Introduction

Osteoarthritis (OA) is a common joint disorder caused by 'wear and tear' on a joint. Knee cartilage deteriorates as people age, eventually leading to a complete loss of cartilage in the joint. When cartilage completely breaks down, bones rub against each other, resulting in pain, swelling, and stiffness in the joint (2011). In 2003, an estimated 45.8 million adults had doctor diagnosed arthritis, and by 2030 this number is projected to increase to 67 million (Hootman and Helmick, 2006). For OA patients, total knee arthroplasty (TKA) is a safe and cost-effective treatment for alleviating pain and restoring physical function in patients who do not respond to nonsurgical therapies. Currently, 500,000 TKA surgeries take place annually in the US (Kurtz et al., 2007). By 2030, this number is expected to increase to over 3.48 - 4.3 million annually(Kurtz et al., 2007; Kurtz et al., 2009), and many of these patients will be under the age of 65 (Hootman and Helmick, 2006).

There are many possible outcomes for a TKA. A successful surgery is considered to be one in which there is improvement over the pre-operative condition. However, sometimes these "successful" surgeries produce suboptimal results, meaning that patients have difficulty performing important activities of daily living. Research suggests that 50% of patients are unable to climb steps (Byrne et al., 2002) and 75% of patients have difficulty squatting (Weiss et al., 2002). Some TKAs are so unsuccessful that a TKA revision (or repeated surgery) needs to take place.

What causes the large variation in the outcomes of TKA? There are several factors that determine the success of a TKA, including proper patient selection, prosthesis design, the preoperative condition of the joint, surgical technique, and postoperative rehabilitation (Stulberg et al., 2002). Studies have found that the most common cause for revision TKA is error in surgical technique because it is highly variable (Stulberg et al., 2002). It is difficult to accurately determine the correct location of crucial alignment landmarks that are used to place the prosthetic components (Siston et al., 2007; Siston et al., 2006; Siston et al., 2005). Furthermore, the accuracy of the limb and implant alignment and stability is confirmed based upon a visual inspection conducted at the conclusion of the replacement procedure. Improper implant positioning and alignment can lead to accelerated wear and loosening and suboptimal functional performance (Stulberg et al., 2002).

Although many patients report relief of pain and improved quality of life following TKA surgery, the knee joint function is never completely restored to that of a native knee (Byrne et al., 2002). Often times, patients find that they have difficulty performing many daily living activities, such as gardening. A survey of 176 TKA patients at least one year after surgery, found that as a result of their knee replacement 75% had difficultly squatting, 72% had difficulty kneeling and 54% had trouble gardening (Weiss et al., 2002). It was also found that kneeling and gardening were activities highly valued by the survey participants (Weiss et al., 2002).

Studies have found that the alignment of the prosthesis greatly affects the post-operative functionality of the implant (Bäthis et al., 2004; Jeffery et al., 1991). Proper implant positioning and alignment can prolong the life of the replacement and ensure proper functionality (Stulberg et al., 2002). Improper positioning can cause instability and loosening which can lead to revision TKA (Bäthis et al., 2004). Therefore, determining the ranges of alignment and positioning of the prosthetic components that produce “optimal” functionality following surgery may help orthopaedic surgeons to ensure a successful postoperative outcome for their patients.

Although using experiments to understand movement dynamics is a good method, it poses two major limitations. First, in experiments it can be difficult to measure important variables, such as muscle, ligament, and joint contact forces (Delp et al., 2007; Piazza, 2006). Second, from experimental data alone it is difficult to determine cause-effect relationships in complex dynamic systems (Delp et al., 2007). Dynamic simulations can be used to complement experiments because they provide information that is hard to acquire experimentally. For example, simulations can provide estimates of muscle and joint forces. Furthermore, simulations enable cause-effect relationships to be identified and allow “what if?" studies to be performed (Delp et al., 2007). In the case of a TKA, the effect of varying the orientation of tibial and femoral components on the forces and kinematics of the prosthetic knee can be determined through the use of a forward simulation of the Oxford Rig.

1.1 Focus of Thesis

This project focuses on the alignment of one specific type of TKA implant, a posterior substituting (PS) implant. With a PS implant, the posterior cruciate ligament (PCL) is sacrificed and replaced with a tibial post and femoral cam. The purpose of this project is to determine how variability in surgical alignment of the femoral and tibial components of a PS implant impacts the biomechanics of the knee during a simulated squatting motion. A forward dynamic simulation of the Oxford Rig was used to run the simulations, varying the components in the frontal, transverse, and sagittal planes. Using the posterior substituting version of the Scorpio implant from Stryker Orthopaedics, the effects of component alignment on ligament and quadriceps muscle forces as well as knee kinematics were analyzed.

1.2 Significance of Research

The number of TKA surgeries taking place annually is rising and the age of patients receiving TKA is decreasing. As a result, the need for better performance functionality following TKA surgery is increasing in need as patients need the ability to return to high-demand physical activities (such as running) that are important to them. Prosthetic alignment has been shown to greatly affect the functionally of the implant postoperatively (Bäthis et al., 2004; Jeffery et al., 1991). A computer simulation is a very useful tool when determining the relationship between component alignment and biomechanical parameters. The computer simulation enables the alignment to be systematically varied and run several times so that the relationship between alignment and the parameters can be determined. This research determined the rotational alignments of the tibial and femoral components that have a significant effect on kinematics, ligament forces, and quadriceps forces during a simulated squatting motion. Understanding which alignments have the most significant effect on biomechanical parameters will provide a rational basis for the alignments that surgeons should focus on achieving during surgery. In the future, this will hopefully allow for a better functional performance for patients following TKA surgeries.

1.3 Overview of Thesis

This thesis has three successive chapters. Chapter 2 describes the Oxford Rig and the simulations that were run varying the rotational alignment of the tibial and femoral components of the PS implant. Chapter 3 discusses the results from the simulations and the effect the alignment has on the different biomechanical parameters. The final chapter, chapter 4, contains the conclusion which presents the key contributions, future applications and directions of this research.

Chapter 2: Methods

2.1 Introduction

There is a significant amount of variation in the alignment of the tibial and femoral components in TKA patients. Poor alignment of these components has been linked to causing suboptimal mechanical outcomes. This study was conducted to gain a better understanding of the relationship between component alignment and knee mechanics during a simulated squatting motion. This was done using similar methods and the same forward dynamic simulation of the Oxford Rig device that was used by Thompson et al. (2011). Building off of Thompson's work, variations in component alignment was examined in not only the transverse plane, but also in the frontal and sagittal planes, and in combinations (Thompson et al.).

2.2 The Oxford Rig and Oxford Rig Simulation

Forward dynamic simulations are simulations in which motion is produced through forces. This is done by differentiating equations of motion and numerically integrating them forward in time subject to gravity, inertial and velocity-dependent effects, and muscle forces (Piazza, 2006). The Oxford Rig, shown in Figure 1, is used to simulate flexed-knee stance, which occurs for example, when riding a bicycle, climbing stairs, or rising from chairs (Zavatsky, 1997). The Rig allows the knee to move in its natural six degrees-of-freedom. This means that the knee is capable of three rotations (internal/external, flexion/extension, abduction/adduction) and three translations (medial/lateral, proximal/distal, anterior/posterior) (Zavatsky, 1997).

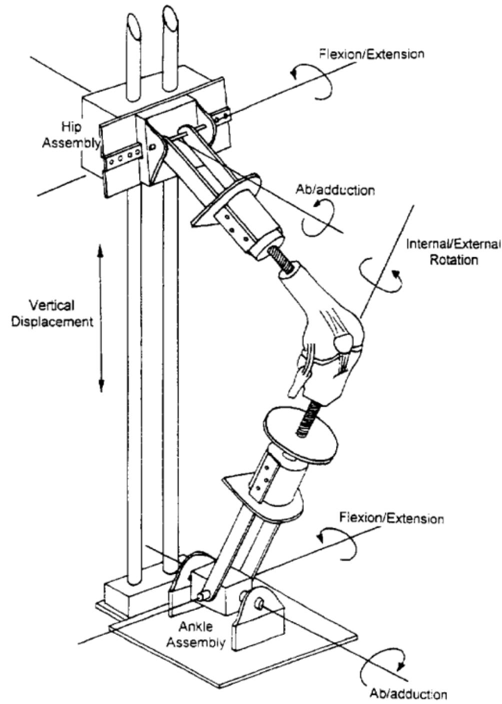

Figure 1. Oxford Rig Assembly (Zavatsky, 1997).

The forward dynamic simulation of the Oxford Rig simulates controlled knee flexion from 20^{o}-120^{o}, as shown in Figure 2. The simulation allows the pelvis to translate vertically directly over the heel which is pinned in place. A proportional derivate controller acts through a lumped quadriceps muscle along the line of the vastus intermedius to produce the force that is required to lower the pelvis through controlled knee flexion. The simulation has a 30 kg mass placed at the pelvis to simulate on half of a person's body weight (Thompson et al., 2011). The LCL, MCL, PCL, and patellar ligament were modeled as springs with quadratic force-deformation relationships (Piazza and Delp, 2001).

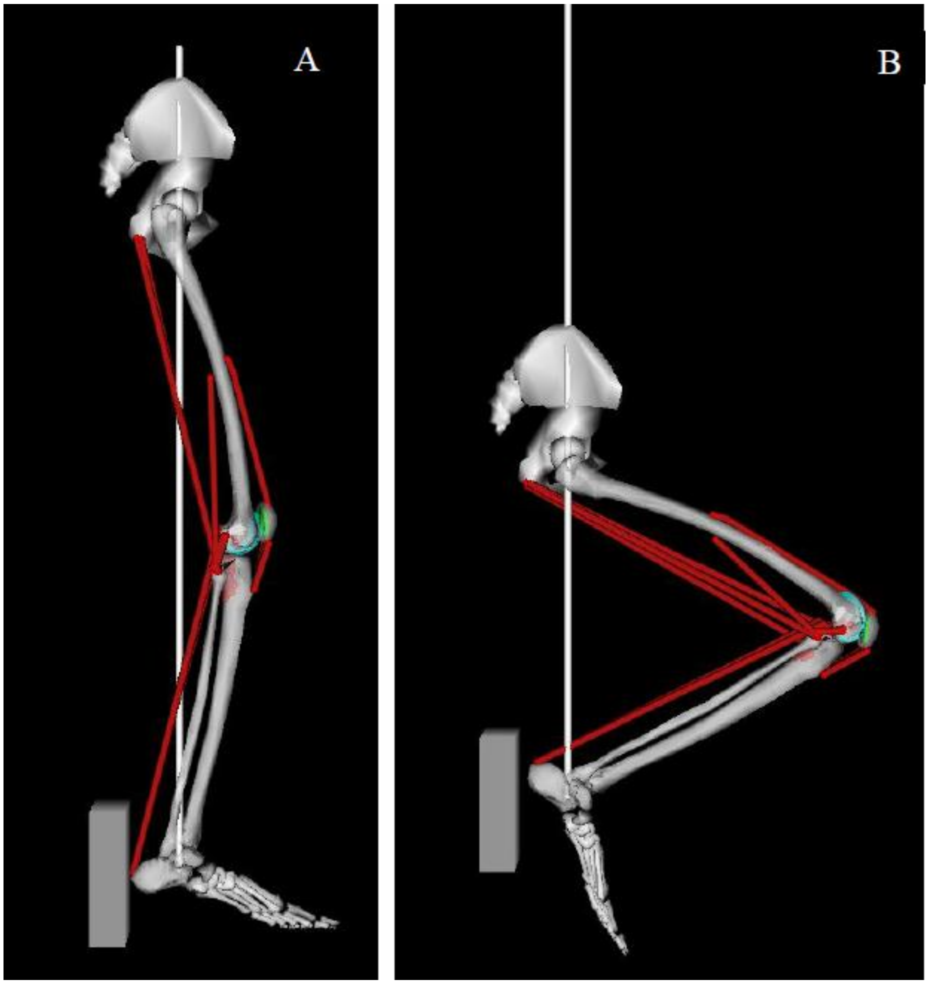

Figure 2. Oxford Rig Computer Simulation Model at (A) 20^{o} of Knee Flexion and at (B) 120^{o} of Knee Flexion.

The dynamic equations of motion were developed using SD/FAST (Symbolic Dynamics, Inc.; Mountain View, CA). The conventions described by Grood and Suntay (1983) were used to determine the knee flexion angles. Using a rigid body spring model, the contact forces in both the tibiofemoral and patellofemoral joint were calculated such that the contact forces depend on the interpenetration of contact surfaces (Landon et al., 2009). A posterior substituting implant (Scorpio CR; Stryker, Inc.) was directly implanted into the simulation. The Oxford Rig simulation, which has been previously validated by Thompson et al. (2011) using a mechanical Oxford Rig device, can be run several times while systematically varying the alignment of the PS prosthetic components while all other parameters are held constant. This will enable the cause-effect relationships between component placement and functional outcome to be determined.

2.3 Simulations

The alignments chosen for the simulations were based on alignments that have been found to be clinically relevant. When aligning in the frontal plane, proper alignment (±3° varus/valgus) to the mechanical axis of the leg is thought to be associated with a better outcome (Bäthis et al., 2004; Jeffery et al., 1991). Studies on postoperative knee alignment have found that alignment that varies more than 3° in both varus and valgus directions, places the component outside the commonly used acceptable range (Bäthis et al., 2004; Chauhan et al., 2004; Sparmann et al., 2003). In the sagittal plane, clinical studies have found that the flexion of the femoral component varies from 0° to 7° and posterior slope of the tibial component varies from -1° to 10° (Chauhan et al., 2004; Sparmann et al., 2003). In the transverse plane, both the femoral and tibial components are associated with high variability in rotational alignment. Studies have found that tibial component alignment can range from 44° internal to 46° external (Siston et al., 2006) and errors in femoral component alignment ranging from 13° internal rotation to 16° external rotation with respect to the transepicondylar axis (Siston et al., 2005).

In this study, a total of 209 simulations were run (see Appendix A for a complete list of all component alignments) to understand the relationship between component orientation and knee mechanics. In the first round of simulations (73 simulations total), only one parameter was varied at a time. Twelve parameters were varied, six for the femoral and tibial components. These parameters included: internal/external rotation, varus/valgus rotation, anterior/posterior slope, anterior/posterior translation, medial/lateral translation and proximal/distal translation. The ranges for the first round of simulations can be found in Table 1.

Table 1. Rotational Parameters Used in the First Round of Simulations.

Alignment Plan	**Range** (Degrees)	**Increment** (Degrees)
Frontal	6/6	2
Sagittal	5/10	2.5
Transverse	15/15	2.5

In the second round of simulations, a total of 136 simulations were run. In this batch of simulations, both the femoral and tibial components were rotated in the same planes. The simulations were run with the ranges and increments shown in Table 2. The simulations performed in the transverse plane were borrowed from Thompson et al. (2011). Using the simulations, the relationship between the rotation of the components and clinical output was determined with linear regression.

Table 2. Rotational Parameters Used in the Second Round of Simulations.

Alignment Plan	**Range** (Degrees)	**Increment** (Degrees)
Frontal	6/6	2
Sagittal	5/10	2.5
Transverse	20/20	5

Chapter 3: Results

3.1 Determining Key Alignments

Linear regression was performed in Minitab®, to determine the relationship between component alignment and biomechanical outcomes by considering linear, quadratic and interaction effects. Component alignments with a regression coefficient greater than an absolute value of 1 and $p \leq 0.05$ were considered to be clinically significant (a regression coefficient greater than an absolute value of 1 implies that ratio between component alignment and the biomechanical parameter is greater than 1:1). The simulation coordinate system is shown in Table 3. The biomechanical parameters of key interest are:

1. Knee kinematics at 20^o and 90^o of knee flexion
2. Lateral collateral ligament (LCL) Force at 20^o and 90^o of knee flexion
3. Medial collateral ligament (MCL) Force at 20^o and 90^o of knee flexion
4. Quadriceps muscle forces at 120^o of knee flexion

Table 3. Simulation Coordinate System.

Component	Transverse Plane		Frontal Plane		Sagittal Plane			
	Internal	External	Varus	Valgus	Anterior Slope	Posterior Slope	Flexor	Recurvatum
Tibial	Positive	Negative	Negative	Positive	Negative	Positive	N/A	N/A
Femoral	Positive	Negative	Positive	Negative	N/A	N/A	Negative	Positive

3.1.1 Knee Kinematics

At 20^o of Knee Flexion

At 20^o of knee flexion, transverse, frontal, and sagittal plane rotation of the tibial and femoral components affects the knee varus/valgus angle differently. When the femoral component is rotated in both the transverse (Figure 3) and frontal planes (Figure 4), the knee varus/valgus will change. This effect is shown through the slope of the line that is created when plotting femoral component rotation versus knee varus/valgus angle. Tibial component rotation in the transverse and frontal planes also affects knee varus/valgus angles. However, rotation of the tibial component in the frontal plane has a larger effect on knee varus/valgus angle than rotation in the transverse plane. This is shown through the small change in knee varus/valgus angle as the tibial component is rotated in the transverse plane (in Figure 3 the vertical displacement between data points is very small). In comparison, when the tibial component is rotated in the frontal plane (Figure 4), there is a larger change in the knee varus/valgus angle (there is a large vertical displacement between data points).

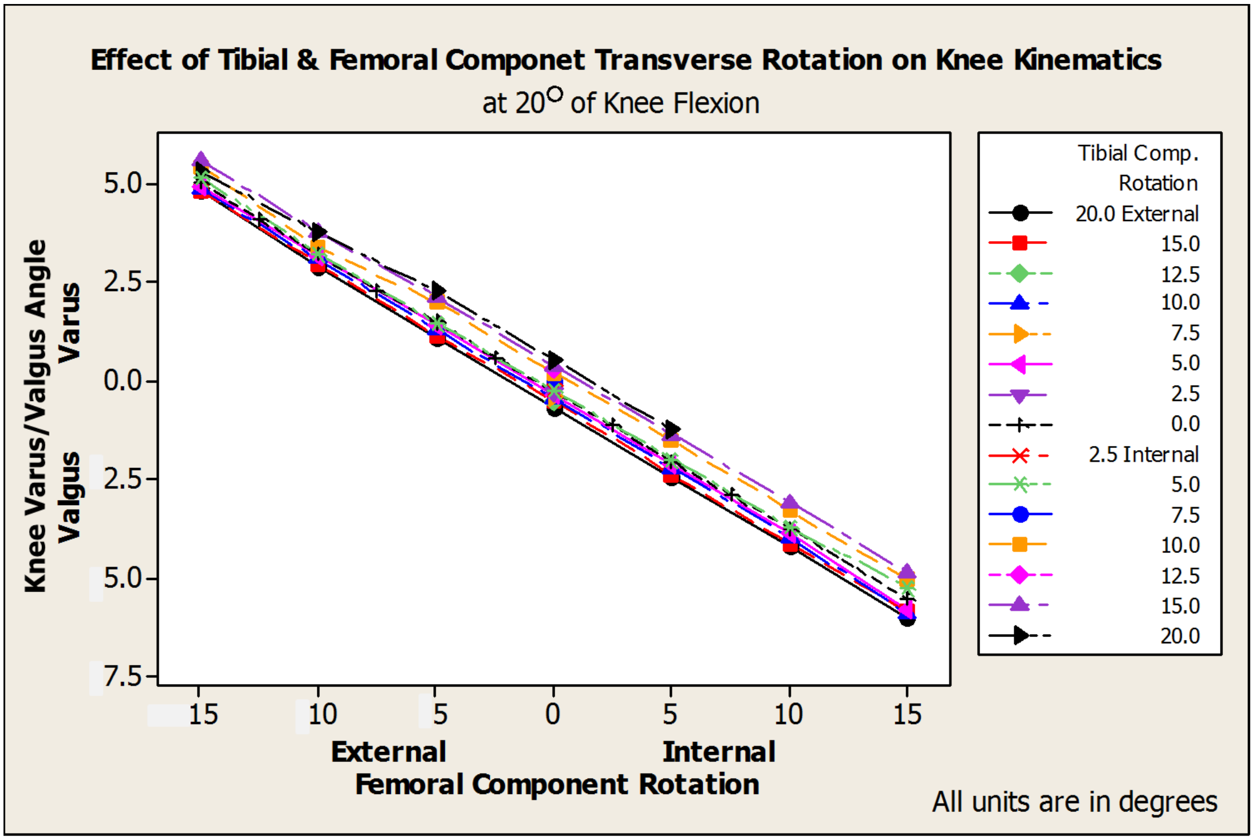

Figure 3. Effect of Tibial and Femoral Component Rotation in Transverse Plane on Knee Kinematics at 20° of Knee Flexion.

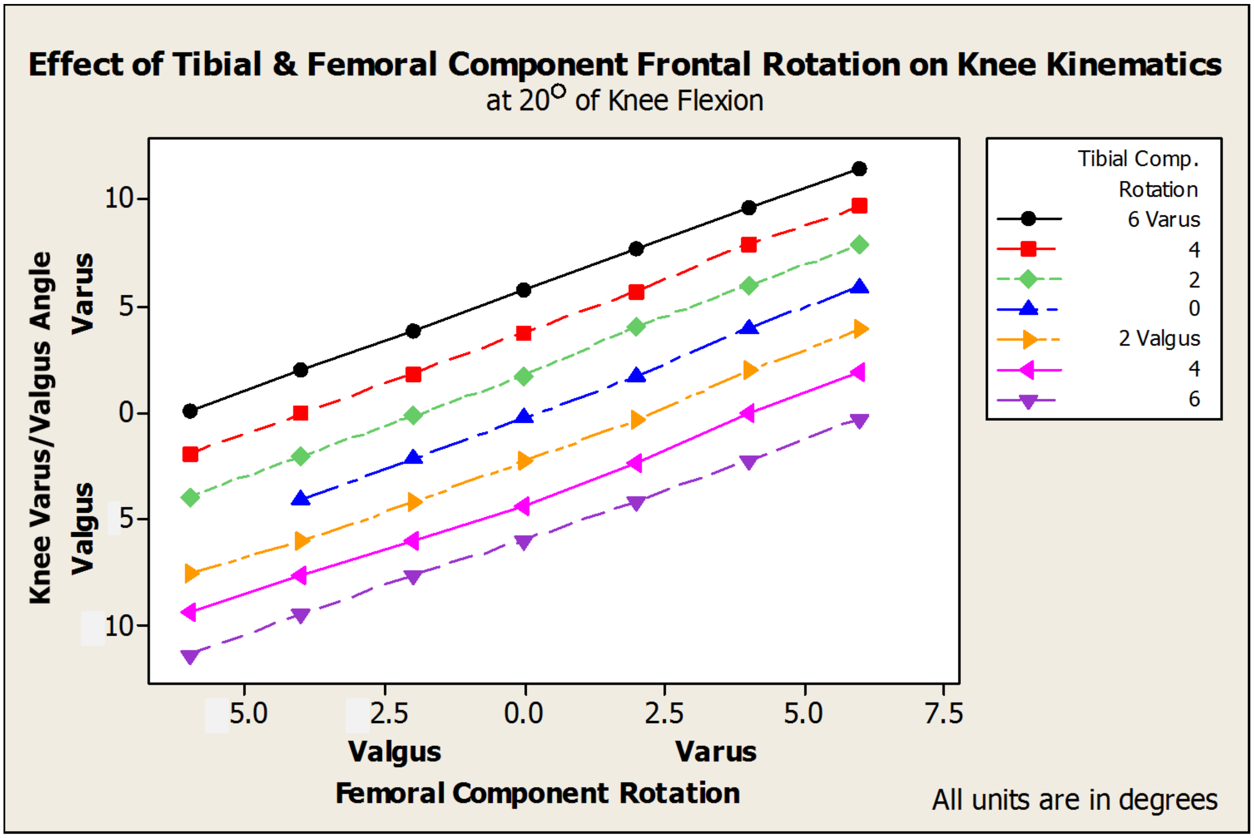

Figure 4.Effect of Tibial and Femoral Component Rotation in Frontal Plane on Knee Kinematics at 20° of Knee Flexion.

When the tibial and femoral components are rotated in the sagittal plane (Figure 5), it has an effect on the knee varus/valgus angle. This effect is very small, as the range of knee angle is about 0.25°. Rotation of the femoral component has very little effect on knee varus/valgus angle.

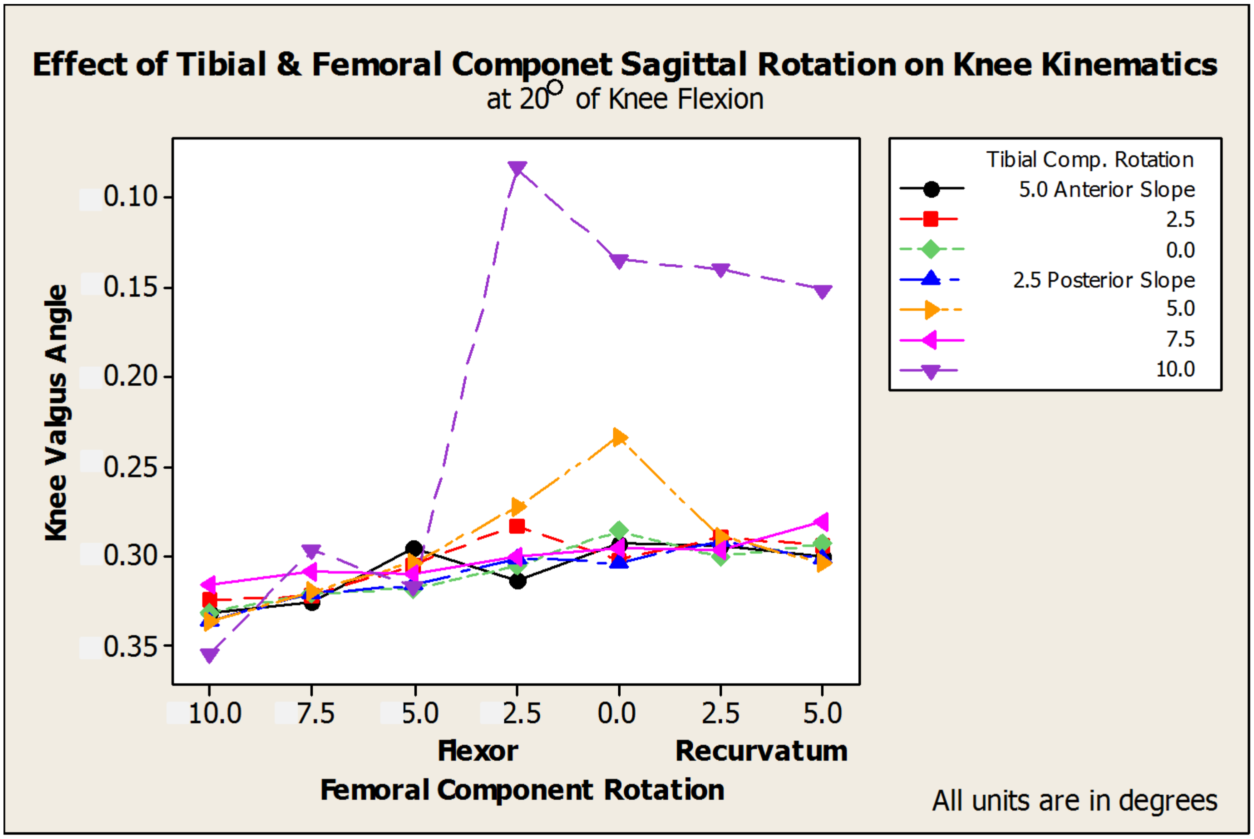

Figure 5. Effect of Tibial and Femoral Component Rotation in Sagittal Plane on Knee Kinematics at 20° of Knee Flexion.

In order to understand how the rotation of the components affects the clinical output, linear regression was performed knee varus/valgus angle at 20° of flexion (Table 4). The largest regression coefficients were found for rotation of both the femoral and tibial components in the frontal plane. The regression coefficients (absolute value) for the femoral and tibial components rotated in the frontal plane were 0.960166 and 0.974909, respectively. This means that for every degree of rotation in the frontal plane, the knee varus/valgus angle is multiplied by a factor of 0.960166 for femoral component rotation and 0.974909 for tibial component rotation. Therefore, knee varus/valgus angle at 20° of knee flexion is most affected by rotation in the frontal plane.

Table 4. Regression Analysis Results for Knee Kinematics at 20° of Knee Flexion.

Knee Varus/Valgus Angle At 20° of Knee Flexion		
Term	**Coefficient**	**P-Value**
Constant	-0.262374	0.000
fc_V/V	**0.960166**	**0.000**
fc_I/E	-0.349972	0.000
fc_F/R	-0.000462	0.921
tc_V/V	**-0.974909**	**0.000**
tc_I/E	0.029163	0.000
tc_A/P	0.002462	0.596
fc_V/V*fc_V/V	0.008267	0.000
fc_I/E*fc_I/E	-0.000032	0.786
fc_F/R*fc_F/R	-0.000999	0.102
tc_V/V*tc_V/V	0.001959	0.031
tc_I/E*tc_I/E	0.000530	0.000
tc_A/P*tc_A/P	0.000353	0.562
fc_V/V*tc_V/V	-0.004095	0.000
fc_I/E*tc_I/E	0.000571	0.000
fc_F/R*tc_A/P	0.000457	0.429

At 90° of Knee Flexion

Rotation of the components in the transverse, frontal, and sagittal planes affects the knee varus/valgus angle at 90° of knee flexion differently. In the transverse plane (Figure 6), the knee varus/valgus angle changes greatly when the femoral component is rotated (this is shown through the sloping line). However, when the tibial component is rotated, there is very little change in the knee varus/valgus angle (this is shown through the very little vertical displacement between data points in Figure 6).

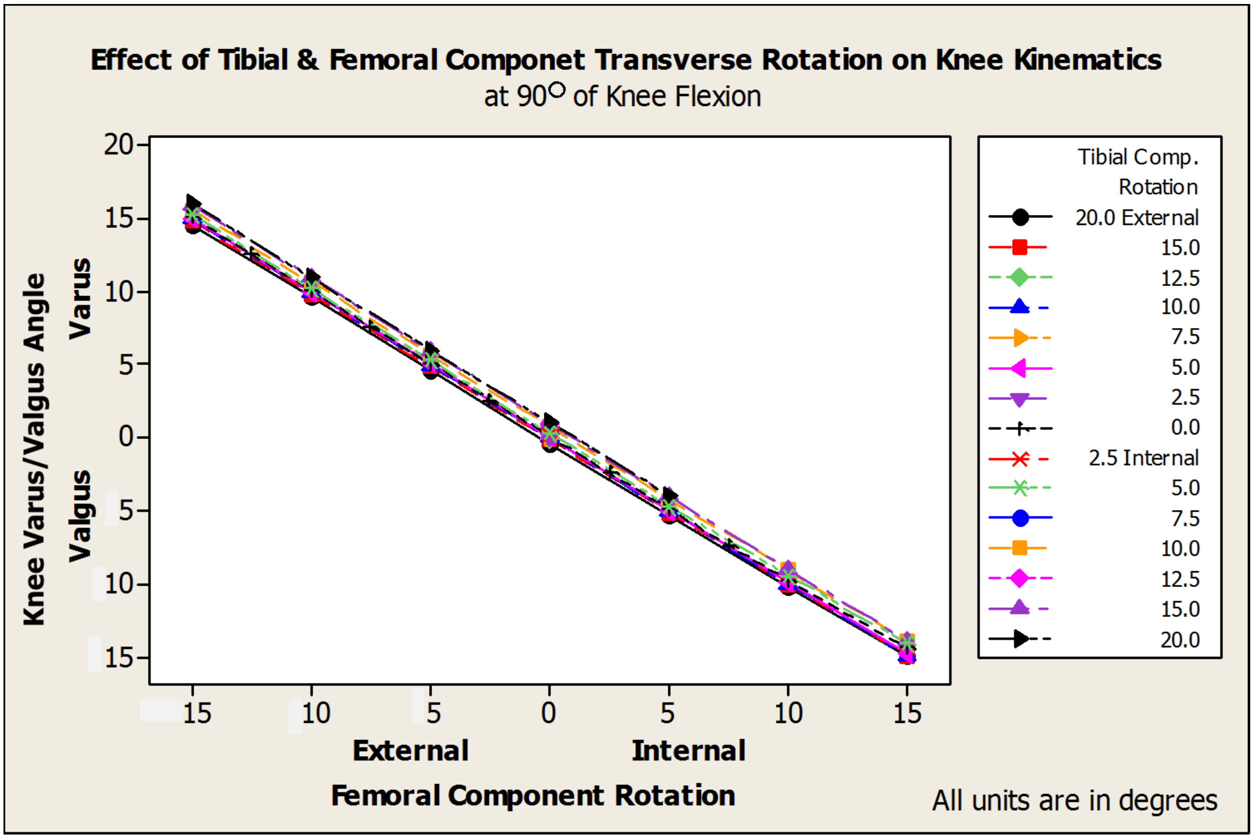

Figure 6. Effect of Tibial and Femoral Component Rotation in Transverse Plane on Knee Kinematics at 90° of Knee Flexion.

In the frontal plane, rotation in the tibial component has a larger effect on knee varus/valgus angle at 90° of knee flexion than femoral component rotation (Figure 7). As shown by the vertical displacement between data points, when the femoral component is rotated the knee varus/valgus angle varies. On the other hand, when the femoral component is rotated, the knee varus/valgus angle remains constant, as shown by the horizontal lines.

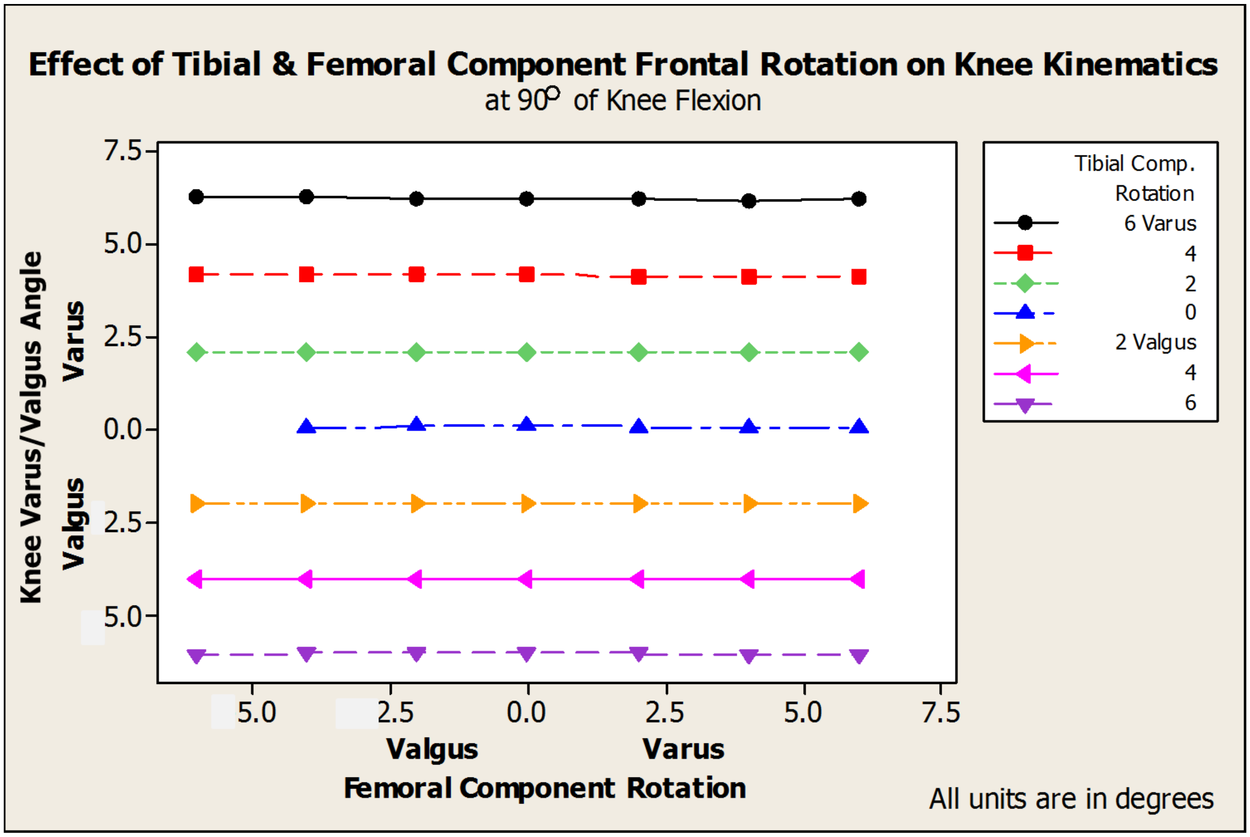

Figure 7. Effect of Tibial and Femoral Component Rotation in Frontal Plane on Knee Kinematics at 90° of Knee Flexion.

Varying the alignment of the femoral and tibial components in the sagittal plane at 90° of knee flexion will cause the knee varus/valgus angle to change over a range of 0.7° (Figure 8).

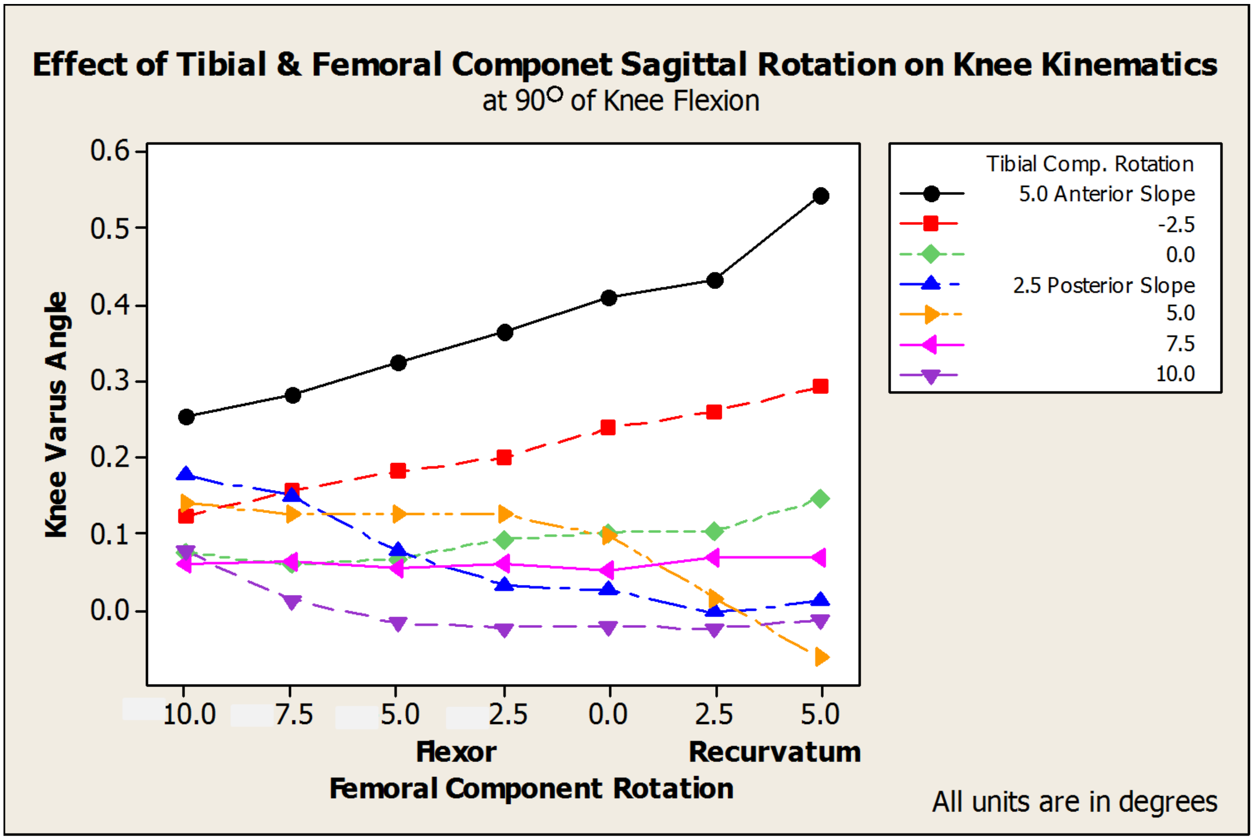

Figure 8. Effect of Tibial and Femoral Component Rotation in Sagittal Plane on Knee Kinematics at 90° of Knee Flexion.

Linear regression analysis was performed in Minitab® to determine the alignments that had significant effect on clinical output (Table 5). Clinical output would be most affected by terms with the largest regression coefficients and $p \leq 0.05$. Rotation of the tibial component in the frontal plane and the femoral component in the transverse plane had the largest absolute value regression coefficients, 1.02132 and 0.98841 respectively. Therefore, varus/valgus rotation of the tibial component and internal/external rotation of the femoral component will have the largest effect on knee varus/valgus angle at 90° of knee flexion.

Table 5. Regression Analysis Results for Knee Kinematics at 90° of Knee Flexion.

Knee Varus/Valgus Angle At 90° of Knee Flexion		
Term	**Coefficient**	**P-Value**
Constant	0.15071	0.000
fc_V/V	-0.00226	0.433
fc_I/E	**-0.98841**	**0.000**
fc_F/R	0.00557	0.121
tc_V/V	**-1.02132**	**0.000**
tc_I/E	0.03679	0.000
tc_A/P	-0.03422	0.000
fc_V/V*fc_V/V	-0.00185	0.011
fc_I/E*fc_I/E	0.00074	0.000
fc_F/R*fc_F/R	0.00017	0.711
tc_V/V*tc_V/V	-0.00081	0.250
tc_I/E*tc_I/E	0.00068	0.000
tc_A/P*tc_A/P	0.00196	0.000
fc_V/V*tc_V/V	0.00050	0.473
fc_I/E*tc_I/E	0.00018	0.030
fc_F/R*tc_A/P	-0.00154	0.001

3.1.2 LCL Force

At 20° of Knee Flexion

At 20° of knee flexion, there is an interaction effect between the tibial and femoral components when rotated in transverse plane (Figure 9). When the femoral component is rotated externally, LCL force will be affected at certain tibial rotations. Regardless of the tibial component rotation, when the femoral component is rotated internally there is no effect on LCL force.

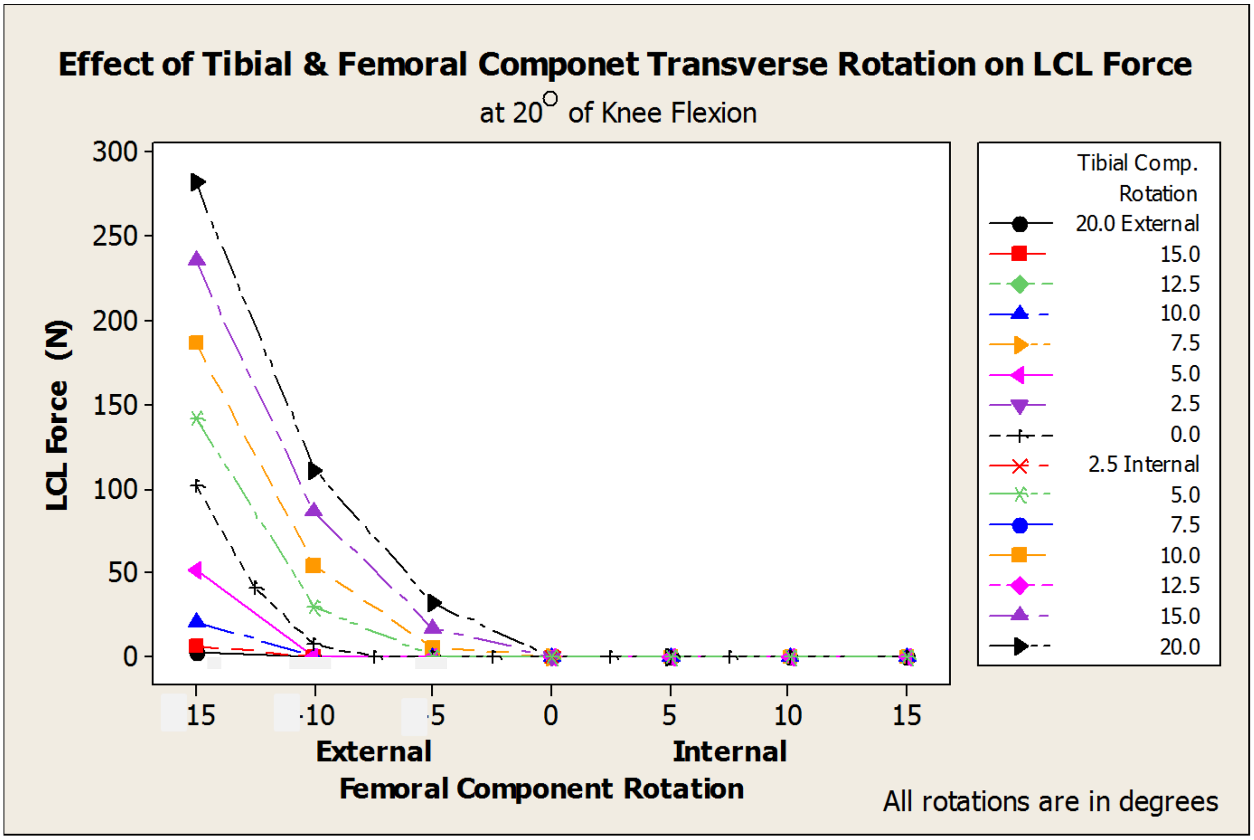

Figure 9. Effect of Tibial and Femoral Component Rotation in Transverse Plane on LCL Force at 20° of Knee Flexion.

There is also an interaction effect between tibial and femoral component alignment in the frontal plane on LCL force (Figure 10).

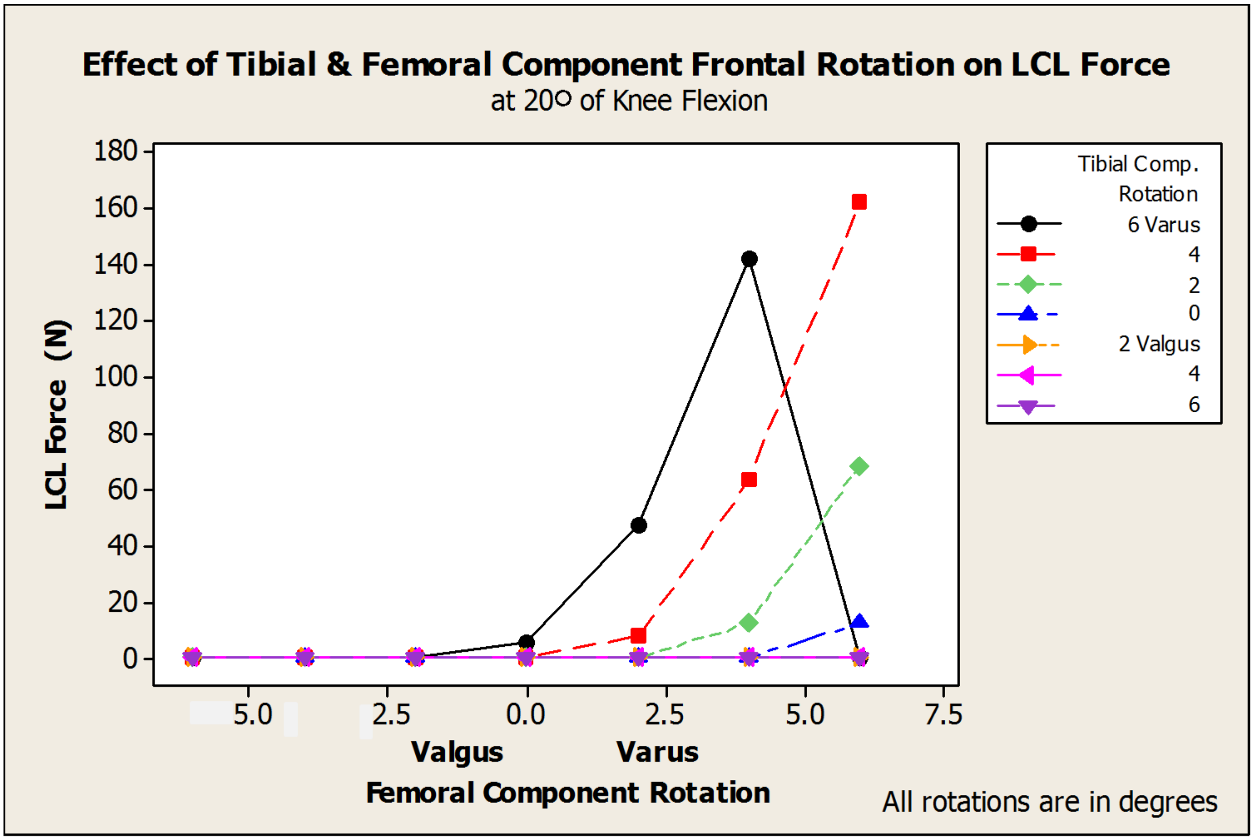

Figure 10. Effect of Tibial and Femoral Component Rotation in Frontal Plane on LCL Force at 20° of Knee Flexion.

When the tibial and femoral components are rotated in the sagittal plane at 20° of knee flexion, LCL force remains constant (Figure 11).

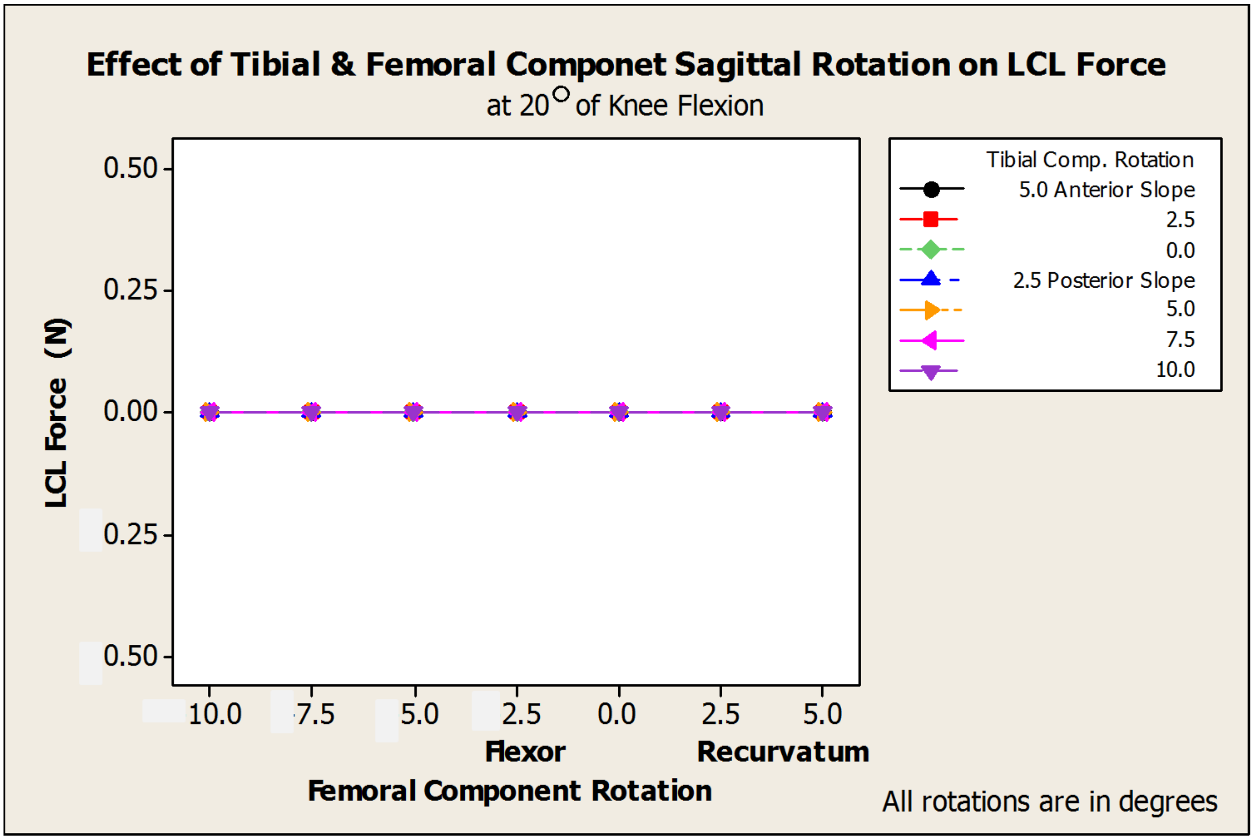

Figure 11. Effect of Tibial and Femoral Component Rotation in Sagittal Plane on LCL Force at 20° of Knee Flexion.

Using Minitab®, linear regression performed to determine the relationship between component alignment and LCL force at 20° of knee flexion (Table 6). Clinical output would be most affected by terms with the largest regression coefficients and p ≤ 0.05. It was found that rotating both the tibial and femoral components in the transverse and frontal planes have significant effect on the LCL force produced at 20° of knee flexion. However, the regression coefficient was largest for femoral component rotation in the frontal plane. Therefore, rotating the femoral component in the frontal plane has the largest effect on LCL force at 20° of knee flexion.

Table 6. Regression Analysis Results for LCL Force at 20° of Knee Flexion.

LCL Force (N) At 20° of Knee Flexion		
Term	**Coefficient**	**P-Value**
Constant	-4.47473	0.115
fc_V/V	**3.15906**	**0.000**
fc_I/E	**-2.87715**	**0.000**
fc_F/R	0.10959	0.911
tc_V/V	**-2.89929**	**0.000**
tc_I/E	**1.32705**	**0.000**
tc_A/P	-0.10959	0.911
fc_V/V*fc_V/V	0.50283	0.011
fc_I/E*fc_I/E	0.22721	0.000
fc_F/R*fc_F/R	0.05845	0.650
tc_V/V*tc_V/V	0.37921	0.049
tc_I/E*tc_I/E	0.02367	0.144
tc_A/P*tc_A/P	0.05845	0.650
fc_V/V*tc_V/V	-0.78836	0.000
fc_I/E*tc_I/E	-0.21395	0.000
fc_F/R*tc_A/P	0.01461	0.905

3.1.3 MCL Force

At 20° of Knee Flexion

At 20° of knee flexion, there is an interaction effect between transverse rotation of the tibial and femoral components and MCL force (Figure 12). Regardless of the alignment of the tibial component, when the femoral component is aligned externally there is no effect on MCL force. As the femoral component is rotated internally, MCL force is affected at certain tibial alignments.

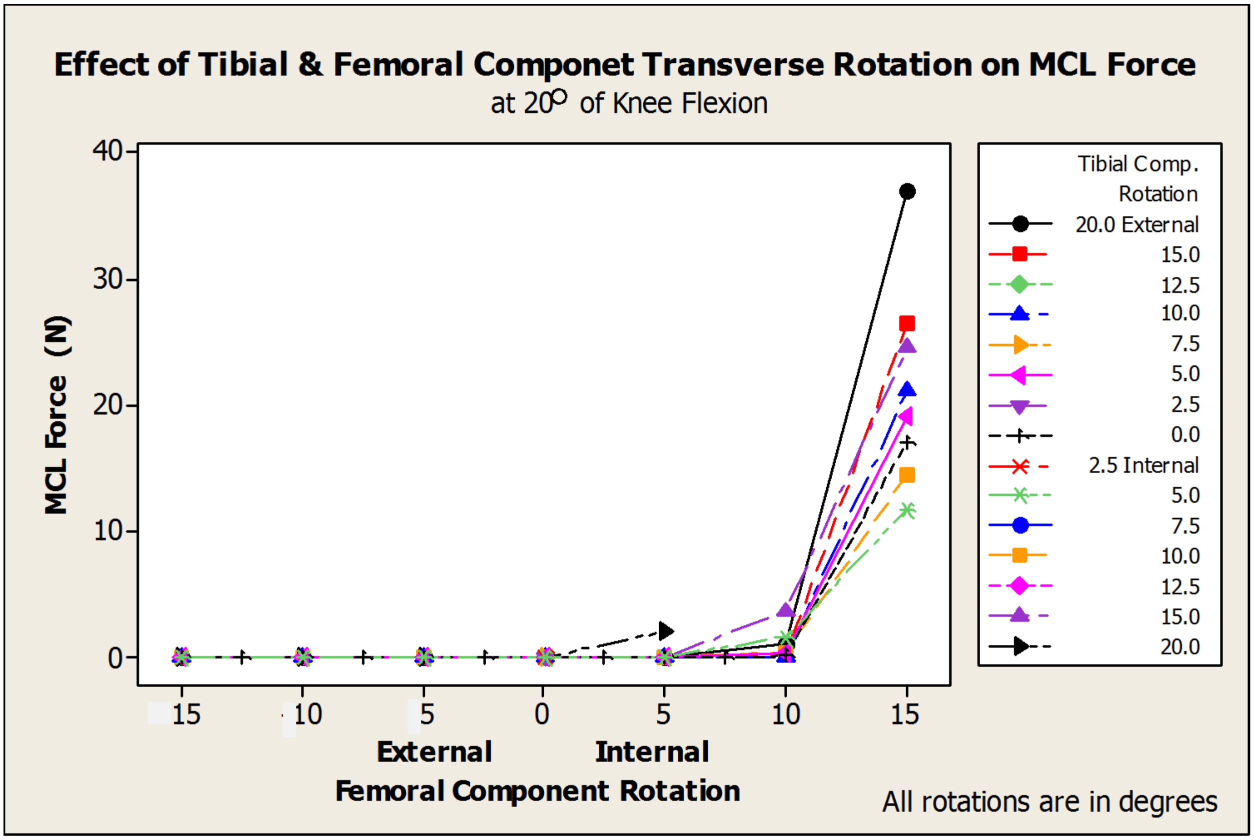

Figure 12. Effect of Tibial and Femoral Component Rotation in Transverse Plane on MCL Force at 20° of Knee Flexion.

There is also an interaction effect between the tibial and femoral components when they are rotated in the frontal plane (Figure 13). As both the tibial and femoral components are rotated in valgus, the MCL force becomes larger. It should be noted that the MCL force is above the published yield point of 453 N (Kennedy et al., 1976) when both the tibial and femoral components are rotated 6° in valgus.

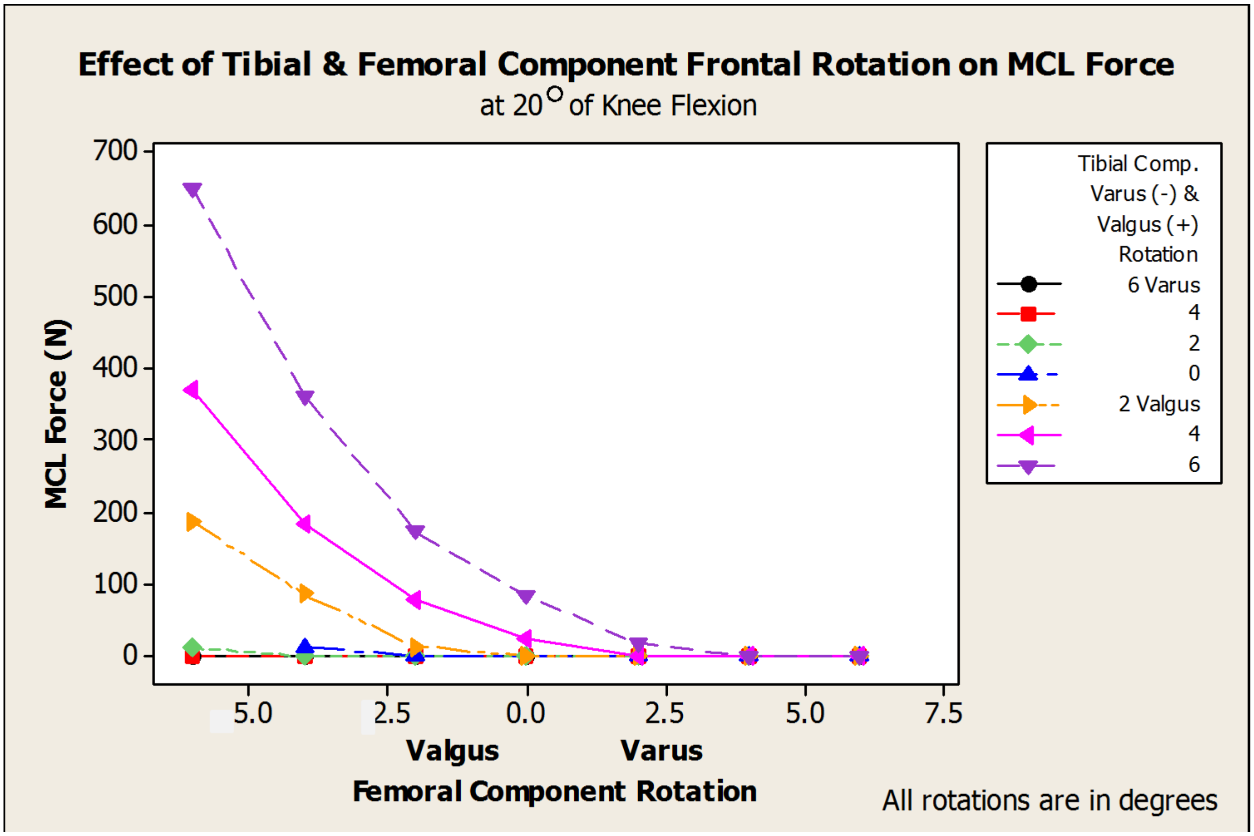

Figure 13. Effect of Tibial and Femoral Component Rotation in Frontal Plane on MCL Force at 20° of Knee Flexion.

Rotating the tibial and femoral components in the sagittal plane has very little effect on the MCL force at 20° of knee flexion (Figure 14). Only when the tibial component is rotated in 10° posterior slope is there an effect on MCL force.

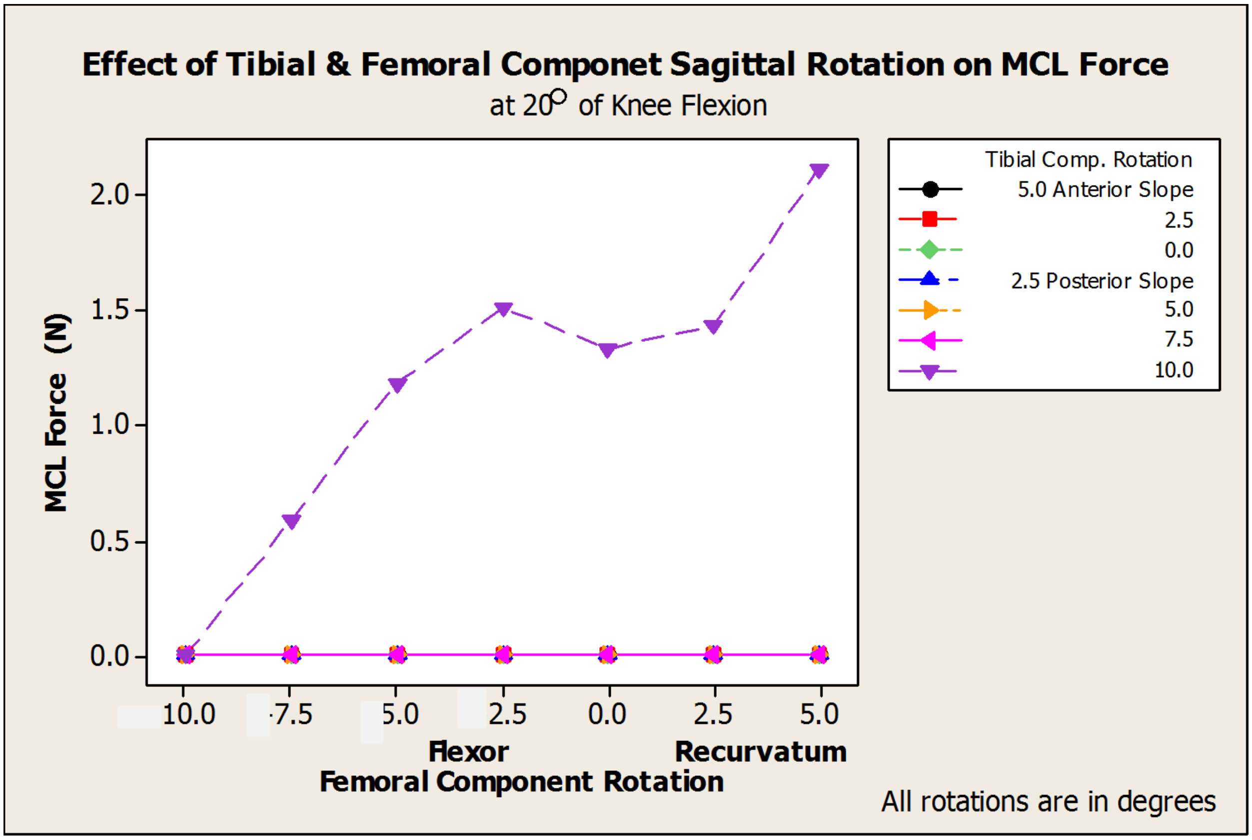

Figure 14. Effect of Tibial and Femoral Component Rotation in Sagittal Plane on MCL Force at 20° of Knee Flexion.

Linear regression was performed in Minitab® to determine which alignments have the biggest effect on MCL force at 20° of knee flexion (Table 7). The alignments with significant effect ($p \le 0.05$ and regression coefficients greatest than absolute value of 1) have been bolded. The regression coefficients for frontal plane rotation for the femoral and tibial components were there largest overall at 14.3976 (absolute value) and 13.8953 respectively. For the femoral component, the coefficient is 14.3976 (absolute value), which means for every 1° of rotation of the femoral component in the transverse plane, the MCL force increases by a magnitude of 14.3976. Likewise, for every 1° of rotation of the tibial component, MCL force increases by a magnitude of 13.8953. Therefore rotating both the tibial and femoral components in the frontal plane will have the most significant clinical effect on MCL force at 20° of knee flexion.

Table 7. Regression Analysis Results for MCL Force at 20^{o} of Knee Flexion.

MCL Force (N) At 20^{o} of Knee Flexion		
Term	**Coefficient**	**P-Value**
Constant	-5.9543	0.098
fc_V/V	**-14.3976**	**0.000**
fc_I/E	0.4656	0.163
fc_F/R	0.1425	0.909
tc_V/V	**13.8953**	**0.000**
tc_I/E	-0.0186	0.944
tc_A/P	-0.1354	0.913
fc_V/V*fc_V/V	**1.9987**	**0.000**
fc_I/E*fc_I/E	0.0618	0.055
fc_F/R*fc_F/R	0.0751	0.645
tc_V/V*tc_V/V	**1.7135**	**0.000**
tc_I/E*tc_I/E	0.0149	0.467
tc_A/P*tc_A/P	0.0870	0.594
fc_V/V*tc_V/V	**-3.9237**	**0.000**
fc_I/E*tc_I/E	-0.0046	0.874
fc_F/R*tc_A/P	0.0240	0.877

At 90^{o} of Knee Flexion

Similar to 20^{o} of knee flexion, at 90^{o} of knee flexion, there is an interaction effect between transverse rotation of the tibial and femoral components and MCL force (Figure 15). Regardless of the alignment of the tibial component, when the femoral component is aligned externally there is no effect on MCL force. As the femoral component is rotated internally, MCL force is affected at certain tibial alignments. When the femoral component is rotated 15^{o} internally, the MCL force is above the published yield point of 453 N (Kennedy et al., 1976).

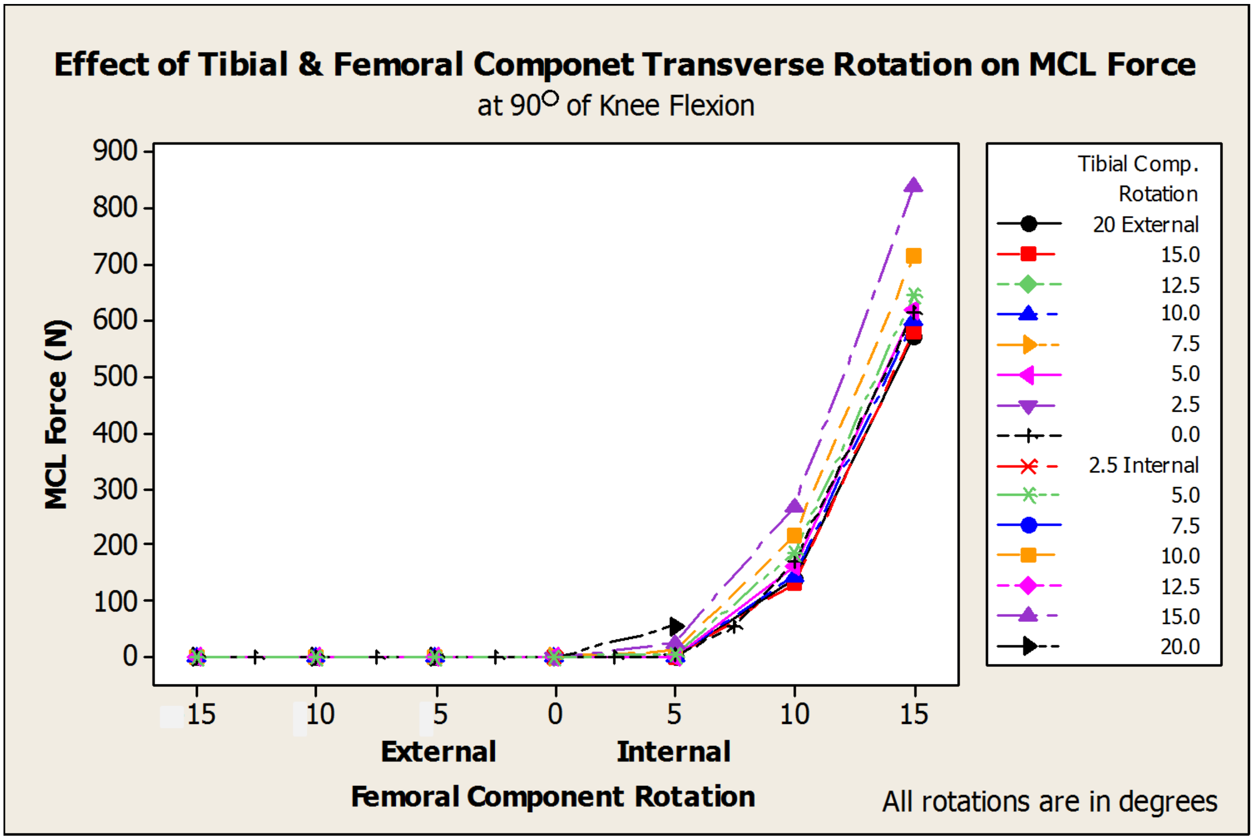

Figure 15. Effect of Tibial and Femoral Component Rotation in Transverse Plane on MCL Force at 90° of Knee Flexion.

In the frontal plane, there is also an interaction effect between the tibial and femoral components on MCL force at 90° of knee flexion (Figure 16). MCL force is only affected when the tibial component is rotated at angles ≥ 2° valgus. Otherwise, regardless of femoral or tibial alignment, MCL force is not produced.

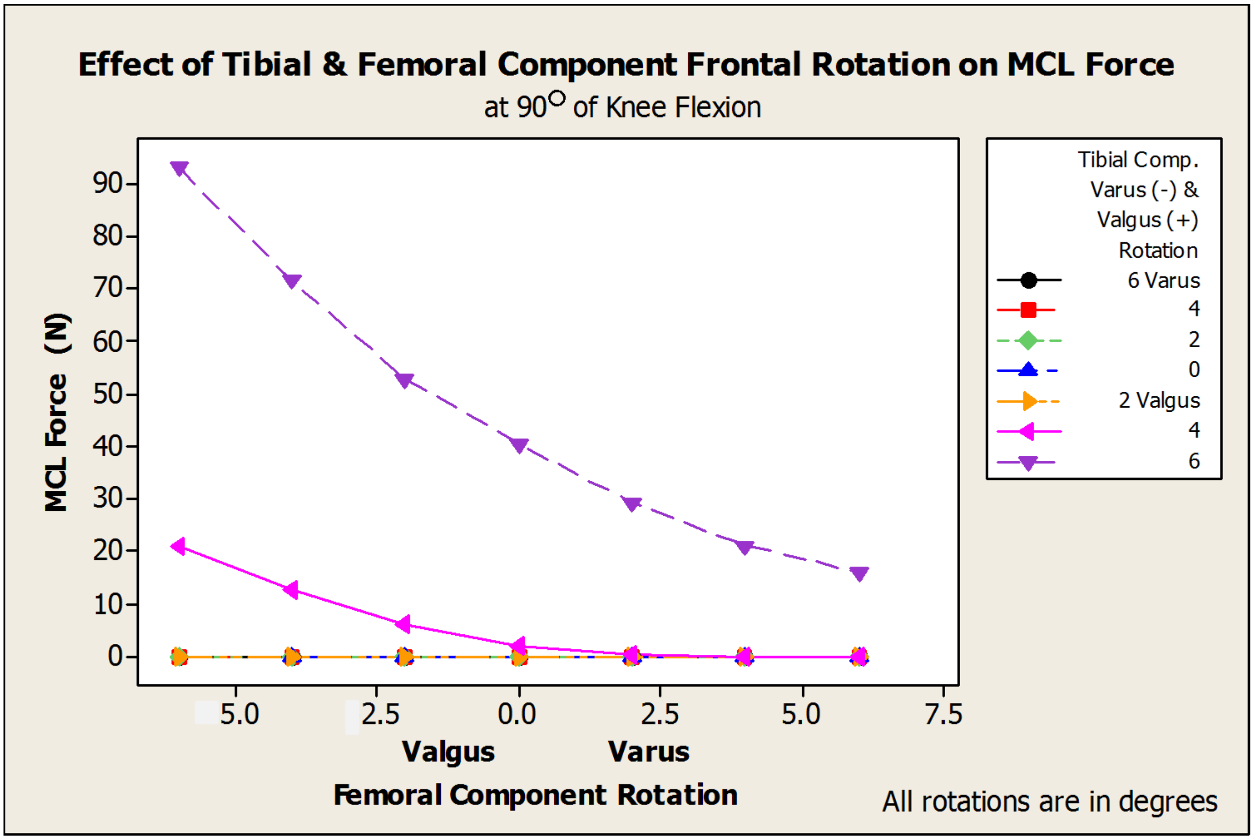

Figure 16. Effect of Tibial and Femoral Component Rotation in Frontal Plane on MCL Force at 90° of Knee Flexion.

When the tibial and femoral components are rotated in the sagittal plane, neither has an effect on the MCL force during 90° of knee flexion (Figure 17).

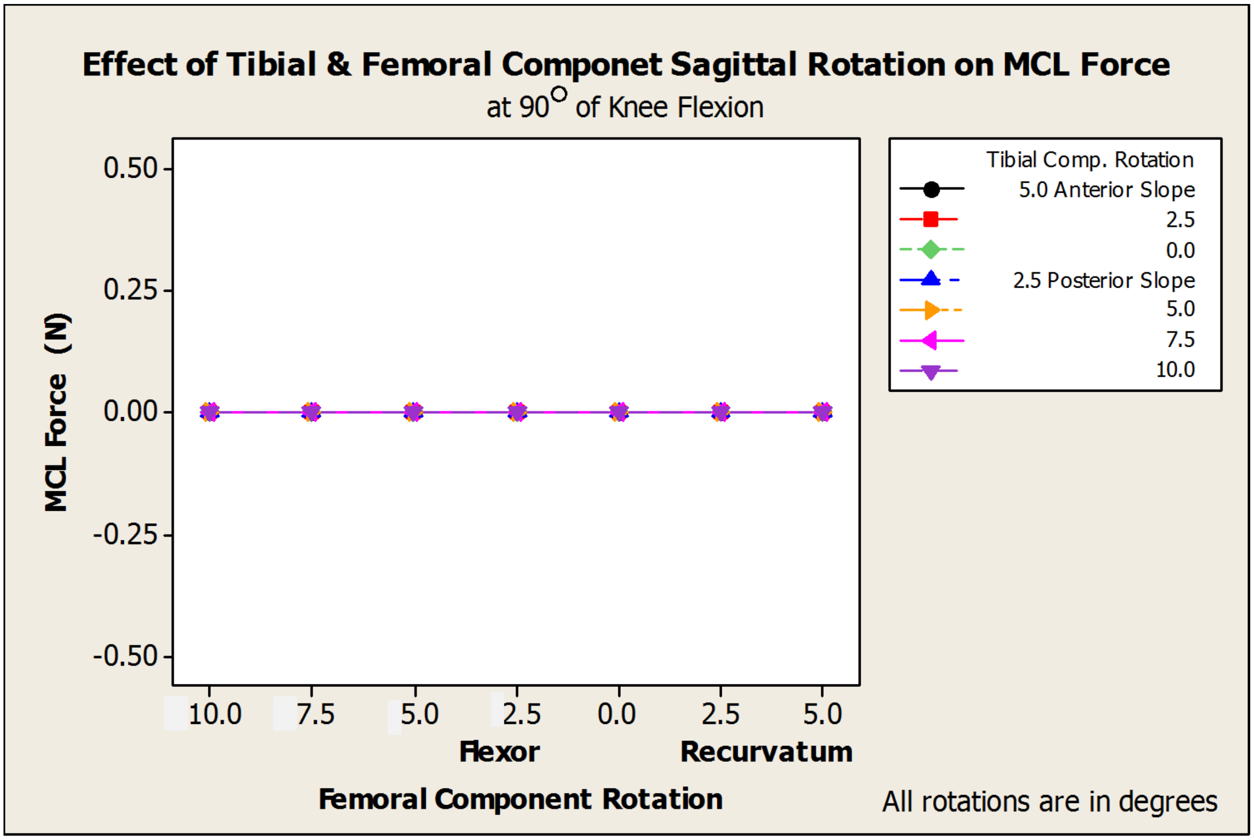

Figure 17. Effect of Tibial and Femoral Component Rotation in Sagittal Plane on MCL Force at 90° of Knee Flexion.

Linear regression was performed to determine what variables have the largest impact on MCL force at 90° of knee flexion (Table 8). The alignments with significant effect ($p \leq 0.05$ and regression coefficients greatest than absolute value of 1) have been bolded. Tibial component alignment in the transverse plane produced a regression coefficient of 1.23894 and femoral component alignment had a regression coefficient of 16.5746. This means that for every 1° of transverse rotation of the tibial and femoral components the MCL force will increase by a magnitude of 1.23894 and 16.5746 respectively. Therefore, MCL force at 90° of flexion is most affected by femoral component rotation.

Table 8. Regression Analysis Results for MCL Force at 90^{o} of Knee Flexion.

MCL Force (N) At 90^{o} of Knee Flexion		
Term	**Coefficient**	**P-Value**
Constant	-13.5463	0.046
fc_V/V	-1.1765	0.532
fc_I/E	**16.5746**	**0.000**
fc_F/R	0.3317	0.887
tc_V/V	2.6886	0.143
tc_I/E	**1.2894**	**0.011**
tc_A/P	-0.3317	0.887
fc_V/V*fc_V/V	0.2957	0.527
fc_I/E*fc_I/E	1.4310	0.000
fc_F/R*fc_F/R	0.1769	0.564
tc_V/V*tc_V/V	0.8758	0.056
tc_I/E*tc_I/E	-0.0064	0.868
tc_A/P*tc_A/P	0.1769	0.564
fc_V/V*tc_V/V	-0.3994	0.384
fc_I/E*tc_I/E	0.1649	0.003
fc_F/R*tc_A/P	0.0442	0.879

3.1.4 Quadriceps Force at 120o of Knee Flexion

In 120^{o} of knee flexion, the effects on quadriceps force from rotating the tibial and femoral components in the transverse plane are additive (Figure 18). Rotating both the tibial and the femoral components in the transverse plane has an effect on quadriceps muscle force.

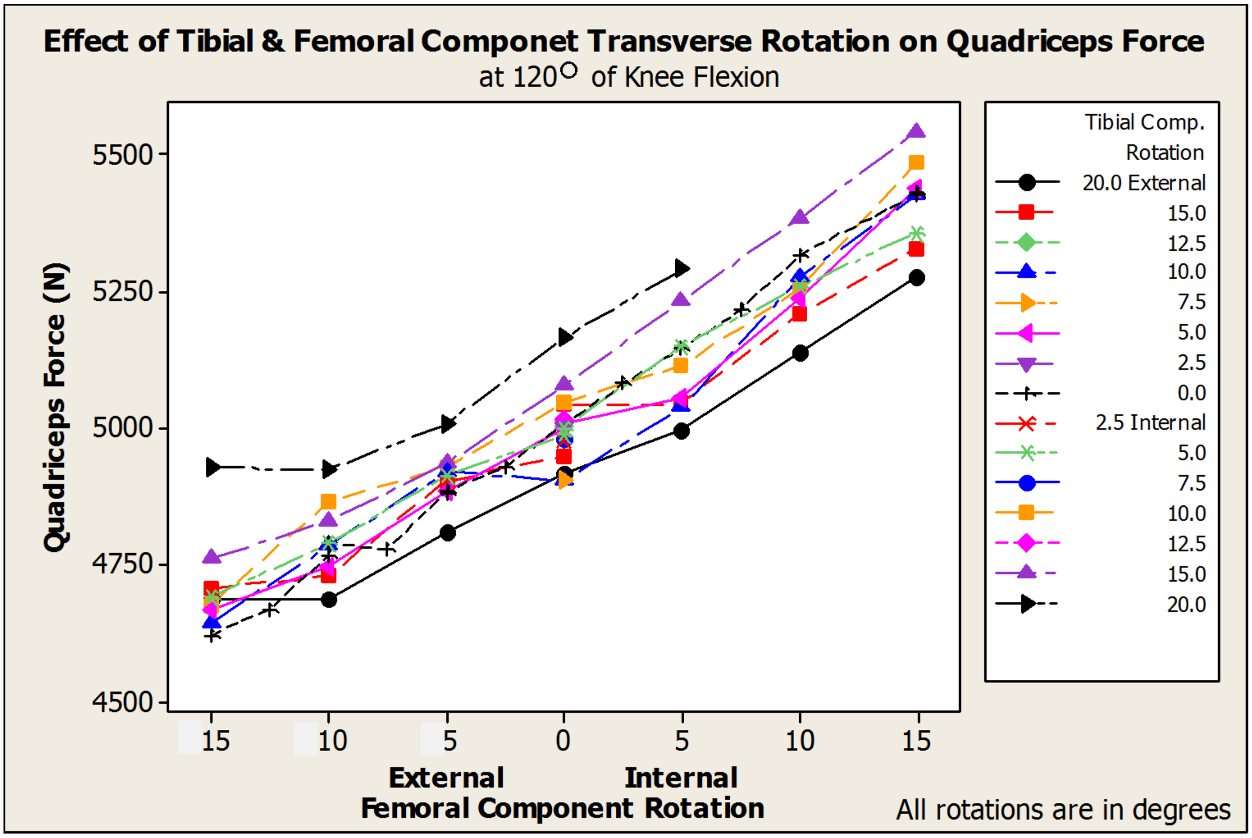

Figure 18. Effect of Tibial and Femoral Component Rotation in Transverse Plane on Quadriceps Force at 120° of Knee Flexion.

Rotating the tibial and femoral components in the frontal plane also has an additive effect on quadriceps muscle force at 120° of knee flexion (Figure 19).

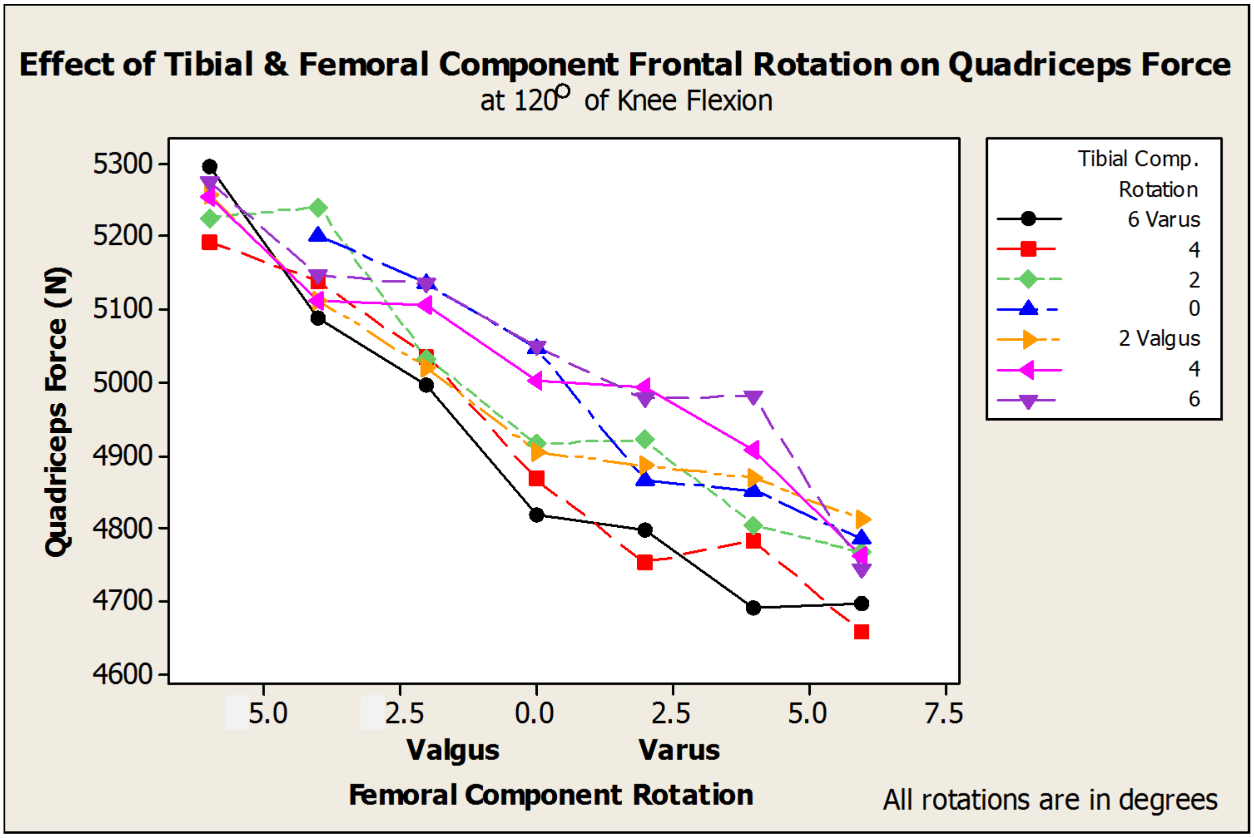

Figure 19. Effect of Tibial and Femoral Component Rotation in Frontal Plane on Quadriceps Force at 120° of Knee Flexion.

Rotating both the femoral and tibial components in the sagittal plane affects the quadriceps (Figure 20).

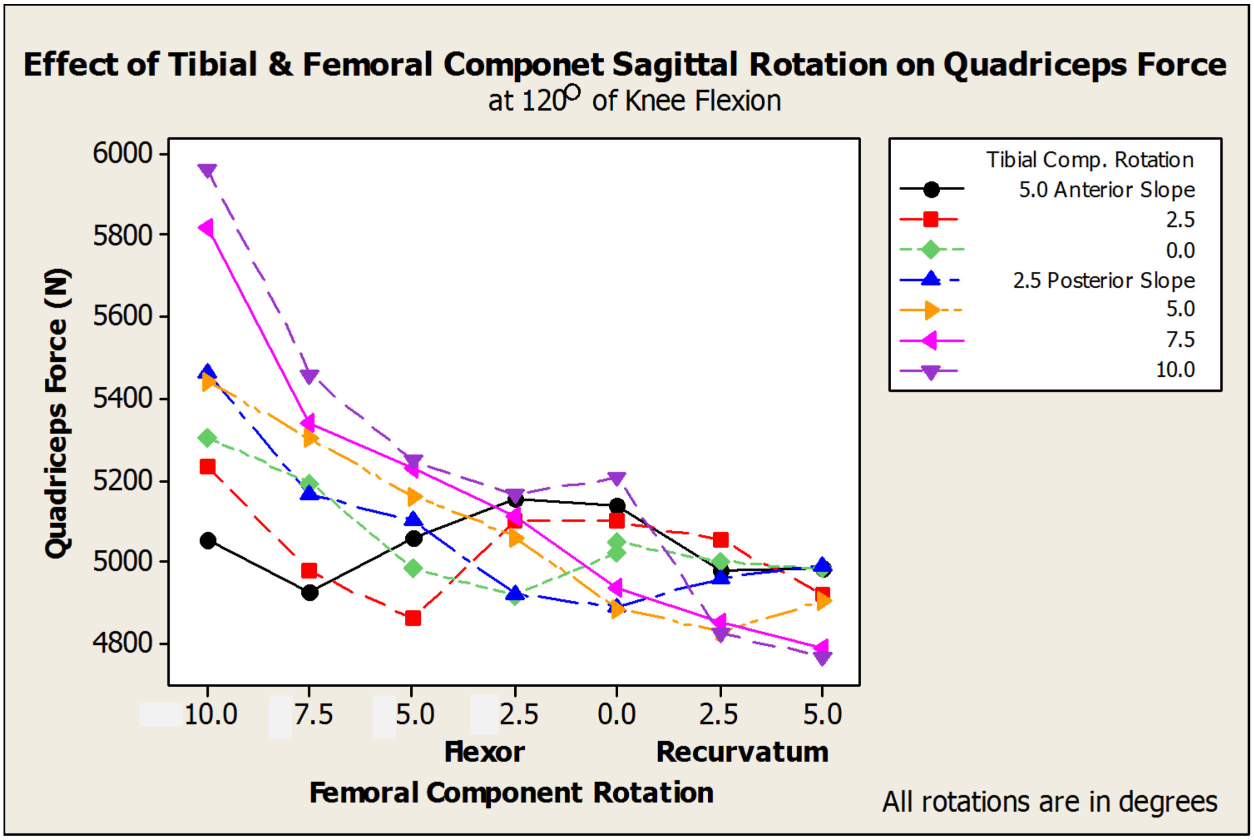

Figure 20. Effect of Tibial and Femoral Component Rotation in Sagittal Plane on Quadriceps Force at 120° of Knee Flexion.

Linear regression was performed in Minitab® to determine which alignments have the biggest effect performed on quadriceps force at 120° of knee flexion (Table 9). The alignments with significant effect ($p \leq 0.05$ and regression coefficients greatest than absolute value of 1) have been bolded. It was found that rotating the alignment of both femoral and tibial components in the frontal, transverse and sagittal planes large effects on quadriceps muscle force. The constant term was found to have the largest regression coefficient (4972.92) followed by femoral component rotation in the transverse plane (41.20 absolute). Therefore, rotating the femoral component in the transverse plane will have the most significant clinical effect on quadriceps force at 120° of knee flexion.

Table 9. Regression Analysis Results for Quadriceps Force at 120° of Knee Flexion.

Quadriceps Force (N) At 120° of Knee Flexion		
Term	**Coefficient**	**P-Value**
Constant	**4972.92**	**0.000**
fc_V/V	**-41.20**	**0.000**
fc_I/E	**24.28**	**0.000**
fc_F/R	**-9.13**	**0.002**
tc_V/V	**10.59**	**0.000**
tc_I/E	**4.89**	**0.000**
tc_A/P	**-5.97**	**0.040**
fc_V/V*fc_V/V	1.13	0.051
fc_I/E*fc_I/E	0.31	0.000
fc_F/R*fc_F/R	2.07	0.000
tc_V/V*tc_V/V	-0.56	0.322
tc_I/E*tc_I/E	0.16	0.001
tc_A/P*tc_A/P	0.94	0.014
fc_V/V*tc_V/V	1.19	0.036
fc_I/E*tc_I/E	0.09	0.199
fc_F/R*tc_A/P	-4.92	0.000

3.1.5 Key Alignments Conclusion

The results from the linear regression are summarized below in Table 10. It contains the alignments that were found to have statistically significant effect on the biomechanical parameters ($p \leq 0.05$ and coefficient greater than 1). The alignments with the largest coefficients are bolded. These bolded alignments are the alignments that have the largest clinical effect. It was found that the biomechanical parameters are most sensitive to alignment variation in the transverse and frontal planes. The femoral component was found to have the most significant effect on LCL force, MCL and quadriceps muscle force but the tibial component has the largest effect on knee varus/valgus angle.

Table 10. Alignment Factors Contributing to Biomechanical Parameters (Principal Alignment in Bold).

Biomechanical Parameters	**Component Alignment Plane**		
	Transverse	**Frontal**	**Sagittal**
Knee Varus/Valgus Angle at 90° of Knee Flexion		**Tibial**	
LCL Force at 20° of Knee Flexion	Femoral, Tibial	**Femoral**, Tibial	
MCL Force at 20° of Knee Flexion		**Femoral**, Tibial	
MCL Force at 90° of Knee Flexion	**Femoral**, Tibial		
Quadriceps Force at 120° of Knee Flexion	Femoral, Tibial	**Femoral** ,Tibial	Femoral, Tibial

3.2 Determining Estimated Tibiofemoral Kinematics Curve

3.2.1 Curve Fitting and Linear Regression

Using the results of the simulations, a model was created to estimate the tibiofemoral kinematics. This was done by finding a second order best fit polynomial for the curve of the kinematics plotted against knee flexion angle for each simulation. The coefficients of each polynomial were then regressed against a constant, the six values (three angles for the femoral and tibial components respectively) that describe the component's rotational alignment and the square of each alignment angle. From this an equation of the form $Ax^2 + Bx + C$, was found to create an estimated curve for the kinematics. For a complete list of coefficients, see Appendix B.

3.2.2 Curve Fitting Results

The results from curve fitting were plotted against the results from the simulations in order to validate that knee kinematics can be assumed to be a second order equation. The estimated knee varus/valgus angle plotted against the simulated knee varus/valgus angle for 20^{o} and 90^{o} of knee flexion are shown in Figure 21 and Figure 22. The result for both 20^{o} and 90^{o} of knee flexion was a linear relationship. Therefore, both quadriceps force and knee kinematics can be modeled by a second order equation. Further, based on the simulation results, the equations found can accurately predict the knee varus/valgus angle when the rotational alignment is known.

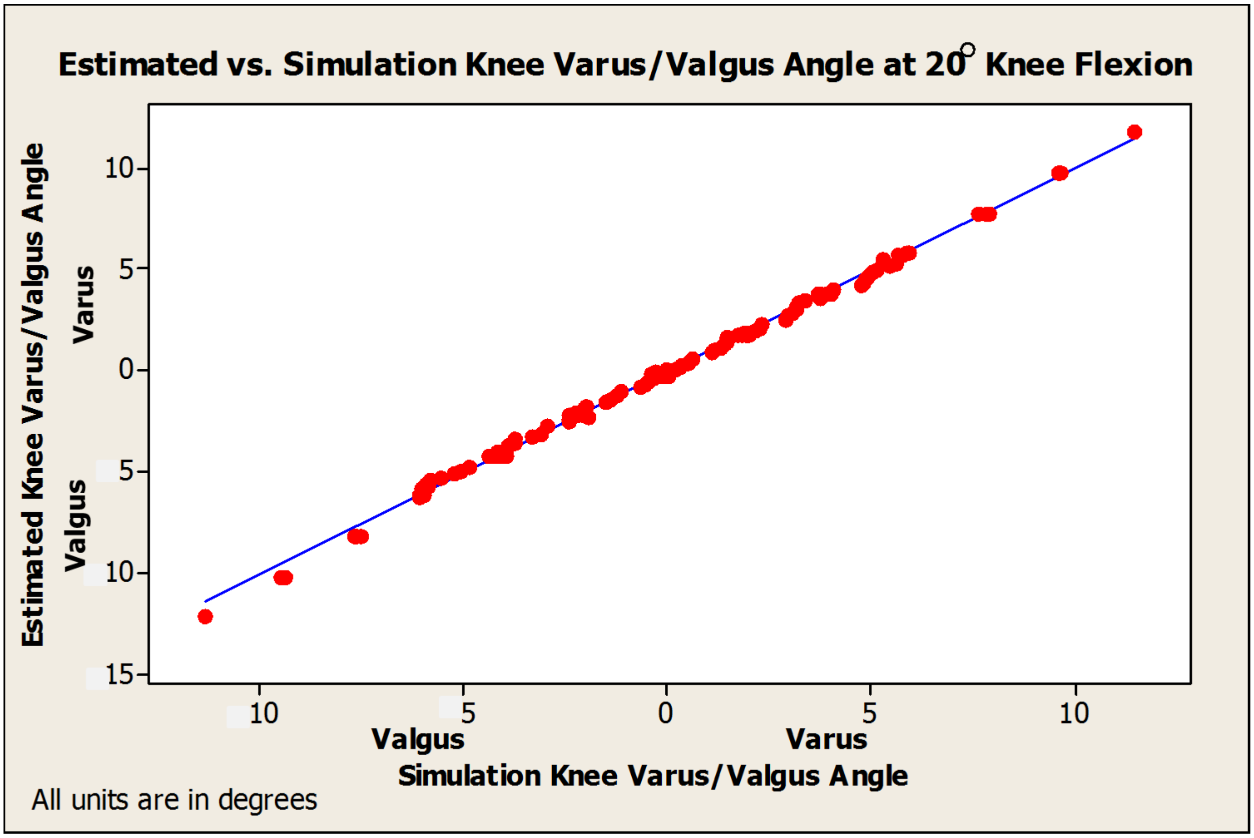

Figure 21. Linear Regression Analysis vs. Simulation Knee Varus/Valgus Angle at 20° of Knee Flexion.

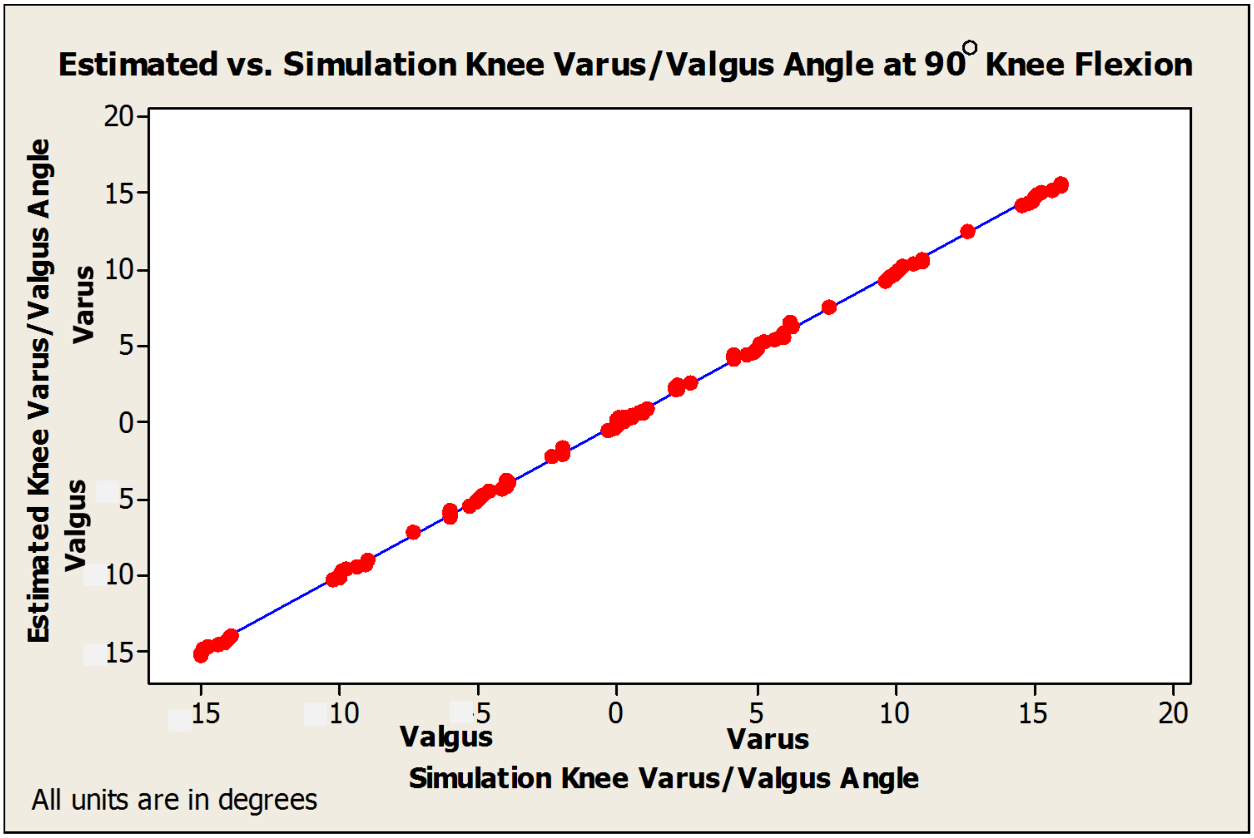

Figure 22. Linear Regression Analysis vs. Simulation Knee Varus/Valgus Angle at 90° of Knee Flexion.

3.3 Discussion

As mentioned previously, alignments in both the transverse and frontal planes for the femoral component are highly variable. In the transverse plane, aligning the femoral component parallel to transepicondylar axis is considered to be the ideal alignment as studies have found this alignment optimizes patellofemoral tracking and minimizes tibiofemoral wear (Miller et al., 2001). This ideal alignment is often hard to achieve due to difficulty in identifying the location of crucial alignment landmarks, such as the medial and lateral epicondyles. In a study conducted by Siston et al. (2005), 11 orthopaedic surgeons found a total of 550 axes to place the femoral component, and only 17.3% of the axes found were within 5° of the transepicondylar axis. In the frontal plane, the ideal alignment is currently a topic of debate. Traditionally, aligning to the line that passes through the center of the knee which connects the center of the femoral head and center of the ankle (the mechanical axis) has been thought to produce the best outcome for TKA patients (Bäthis et al., 2004; Jeffery et al., 1991). Recently however, other research has suggested that restoring the mechanical axis of the knee to its pre-arthritic state, or "naturally aligned" state, will produce a more favorable outcome (Howell et al., 2008).

The results from this study show that with a posterior substituting implant, the MCL, LCL and quadriceps muscle force are most sensitive to changes in femoral alignment in the frontal and transverse planes. Knee kinematics at 90^{o} of knee flexion however, is most sensitive to changes in the tibial component in the frontal plane. Therefore, the alignments that affect the biomechanical outcomes the most in a simulated squatting motion are also the alignments that are most variable. In the transverse plane, the ideal alignment (the transepicondylar axis) is known, but difficult to achieve. In the frontal plane, alignment is easier to achieve than in the transverse plane, but the optimum alignment is in debate.

The Oxford Rig simulation is an idealized squatting motion, which means that it has some important limitations when compared to a true squatting motion. The pelvis is permitted to only translate vertically (all other degrees of freedom have been fixed) and the foot is permitted to plantarflex a large amount as the simulation performs the perfect up-down squatting motion. The only muscle that produces force is the lumped quadriceps muscle and the simulated body weight (the 30 kg mass) is positioned directly above the hip (Thompson et al., 2011). One of the major benefits of this computer simulation is that it allowed for the component orientations to be altered and the effect on the biomechanical parameters to be determined.

Chapter 4: Conclusion

The purpose of this project was to determine the effects of malalignment of the femoral and tibial components of a posterior substituting implant on knee biomechanics during a simulated squatting motion in the Oxford Rig.

4.1 Contributions

Total knee arthroplasty is considered to be a reliable treatment for osteoarthritis of the knee, even though it can produce suboptimal functional outcomes. Proper implant positioning and alignment can prolong the life of the replacement and ensure proper functionality (Stulberg et al., 2002) while malalignment can cause instability and loosening, ultimately leading to a revision TKA (Bäthis et al., 2004). Using a computer simulation of the Oxford Rig, the effect of tibial and femoral component alignments on MCL, LCL, quadriceps muscle force and knee kinematics were studied. The results from this research determined the effects of alignment variation during a simulated squatting motion, which is similar to some of the activities that patients have difficulty performing following a TKA, such as gardening and kneeling (Weiss et al., 2002). After studying a wide range of alignments, it was found that the biomechanical parameters of interest are most sensitive to changes in frontal and transverse plane alignment. This information is a first step in understanding how component alignment affects functional outcomes. This research also determined a model to estimate the tibiofemoral kinematics of the posterior substituting implant. Knowing the alignment of the component, this equation can be used to determine the effect of alignment on knee kinematics.

4.2 Additional Applications

This research determined the effects of varying component alignment in the transverse, frontal, and sagittal planes on biomechanical parameters of interest during a simulated squatting motion. A total of twelve parameters were varied during these simulations. Two different batches of simulations were run; in the first, only one component alignment was varied in a single plane and in the second, the femoral and tibial components were rotated in the same rotational plane. Examining the femoral and tibial components individually is also important. The alignment of each component should be changed in all three planes in order to get a better understanding of how alignment of the tibial and femoral components affects the biomechanical parameters of interest. Further, the simulations can then be used to model ligament force and quadriceps muscle force, so that an estimated curve for the more biomechanical parameters (MLC, LCL, and quadriceps muscle force) can be found.

4.3 Future Work

Using the Oxford Rig as inspiration, additional forward dynamic simulations could be created to simulate other tasks, such as walking, stair climbing, and running. Using these simulations, this research could then be repeated to examine how component alignment affects the knee mechanics when performing these tasks. The results from this thesis could then be compared to the results from the newly developed equations. If there are similarities, than this information could then be used by surgeons to make sure they focus on achieving the key alignments. This could be a key step in achieving the long term goal of helping patients to have higher functionality following TKA surgeries.

4.4 Summary

The biomechanical effects of alignment variations in femoral and tibial component alignment in a simulated squatting motion were determined using a forward-dynamic simulation of the Oxford Rig. It was found that MCL, LCL, and quadriceps muscle force are sensitive to femoral alignment in the transverse and frontal planes. It was also found that knee kinematics is sensitive to tibial alignment in the frontal plane. Further, a model was developed which enabled an equation estimating the curve for knee kinematics to be determined. This research serves as an important first step in determining the alignments of femoral and tibial components that will produce the optimal post-operative functionality for patients. Hopefully, this will one day lead to patients having the ability to participate in high-demand physical activities.

Appendix A

Complete list of simulation coordinate system, variables, and component alignments used in the 209 simulations.

Table 11. Simulation Coordinate System.

Component	Transverse Plane		Frontal Plane		Sagittal Plane			
	Internal	External	Varus	Valgus	Anterior Slope	Posterior Slope	Flexor	Recurvatum
Tibial	Positive	Negative	Negative	Positive	Negative	Positive	N/A	N/A
Femoral	Positive	Negative	Positive	Negative	N/A	N/A	Negative	Positive

Table 12. Definition of Variables Used in Component Alignments.

Variable	Rotation/Translation
Rx	Varus/Valgus Rotation
Ry	Internal/External Rotation
Rz	Tibial Component : Anterior/Posterior Slope Rotation Femoral Component : Flexor/Recurvatum Rotation
Tz	Medial/Lateral Translation
Tx	Anterior/Posterior Translation
Ty	Proximal/Distal Translation

Table 13. Complete List of Simulation Component Alignments.

#	Round #	Femoral Component						Tibial Component					
		Rx	Ry	Rz	Tz	Tx	Ty	Rx	Ry	Rz	Tz	Tx	Ty
1	1	-6	0	0	-0.003	0.004	-0.398	0	0	0	-0.009	-0.01	0.387
2	1	-4	0	0	-0.003	0.004	-0.398	0	0	0	-0.009	-0.01	0.387
3	1	-2	0	0	-0.003	0.004	-0.398	0	0	0	-0.009	-0.01	0.387
4	1	0	-15	0	-0.003	0.004	-0.398	0	0	0	-0.009	-0.01	0.387
5	1	0	-12.5	0	-0.003	0.004	-0.398	0	0	0	-0.009	-0.01	0.387
6	1	0	-10	0	-0.003	0.004	-0.398	0	0	0	-0.009	-0.01	0.387
7	1	0	-7.5	0	-0.003	0.004	-0.398	0	0	0	-0.009	-0.01	0.387
8	1	0	-5	0	-0.003	0.004	-0.398	0	0	0	-0.009	-0.01	0.387
9	1	0	-2.5	0	-0.003	0.004	-0.398	0	0	0	-0.009	-0.01	0.387
10	1	0	0	-10	-0.003	0.004	-0.398	0	0	0	-0.009	-0.01	0.387
11	1	0	0	-7.5	-0.003	0.004	-0.398	0	0	0	-0.009	-0.01	0.387
12	1	0	0	-5	-0.003	0.004	-0.398	0	0	0	-0.009	-0.01	0.387
13	1	0	0	-2.5	-0.003	0.004	-0.398	0	0	0	-0.009	-0.01	0.387
14	1	0	0	0	-0.008	0.004	-0.398	0	0	0	-0.009	-0.01	0.387
15	1	0	0	0	-0.0055	0.004	-0.398	0	0	0	-0.009	-0.01	0.387

#	Round #	Femoral Component						Tibial Component					
		Rx	Ry	Rz	Tz	Tx	Ty	Rx	Ry	Rz	Tz	Tx	Ty
16	1	0	0	0	-0.003	-0.001	-0.398	0	0	0	-0.009	-0.01	0.387
17	1	0	0	0	-0.003	0.0015	-0.398	0	0	0	-0.009	-0.01	0.387
18	1	0	0	0	-0.003	0.004	-0.403	0	0	0	-0.009	-0.01	0.387
19	1	0	0	0	-0.003	0.004	-0.4005	0	0	0	-0.009	-0.01	0.387
20	1	0	0	0	-0.003	0.004	-0.398	-6	0	0	-0.009	-0.01	0.387
21	1	0	0	0	-0.003	0.004	-0.398	-4	0	0	-0.009	-0.01	0.387
22	1	0	0	0	-0.003	0.004	-0.398	-2	0	0	-0.009	-0.01	0.387
23	1	0	0	0	-0.003	0.004	-0.398	0	-15	0	-0.009	-0.01	0.387
24	1	0	0	0	-0.003	0.004	-0.398	0	-12.5	0	-0.009	-0.01	0.387
25	1	0	0	0	-0.003	0.004	-0.398	0	-10	0	-0.009	-0.01	0.387
26	1	0	0	0	-0.003	0.004	-0.398	0	-7.5	0	-0.009	-0.01	0.387
27	1	0	0	0	-0.003	0.004	-0.398	0	-5	0	-0.009	-0.01	0.387
28	1	0	0	0	-0.003	0.004	-0.398	0	-2.5	0	-0.009	-0.01	0.387
29	1	0	0	0	-0.003	0.004	-0.398	0	0	-5	-0.009	-0.01	0.387
30	1	0	0	0	-0.003	0.004	-0.398	0	0	-2.5	-0.009	-0.01	0.387
31	1	0	0	0	-0.003	0.004	-0.398	0	0	0	-0.014	-0.01	0.387
32	1	0	0	0	-0.003	0.004	-0.398	0	0	0	-0.0115	-0.01	0.387
33	1	0	0	0	-0.003	0.004	-0.398	0	0	0	-0.009	-0.015	0.387
34	1	0	0	0	-0.003	0.004	-0.398	0	0	0	-0.009	-0.0125	0.387
35	1	0	0	0	-0.003	0.004	-0.398	0	0	0	-0.009	-0.01	0.382
36	1	0	0	0	-0.003	0.004	-0.398	0	0	0	-0.009	-0.01	0.3845
37	1	0	0	0	-0.003	0.004	-0.398	0	0	0	-0.009	-0.01	0.387
38	1	0	0	0	-0.003	0.004	-0.398	0	0	0	-0.009	-0.01	0.3895
39	1	0	0	0	-0.003	0.004	-0.398	0	0	0	-0.009	-0.01	0.392
40	1	0	0	0	-0.003	0.004	-0.398	0	0	0	-0.009	-0.0075	0.387
41	1	0	0	0	-0.003	0.004	-0.398	0	0	0	-0.009	-0.005	0.387
42	1	0	0	0	-0.003	0.004	-0.398	0	0	0	-0.0065	-0.01	0.387
43	1	0	0	0	-0.003	0.004	-0.398	0	0	0	-0.004	-0.01	0.387
44	1	0	0	0	-0.003	0.004	-0.398	0	0	2.5	-0.009	-0.01	0.387
45	1	0	0	0	-0.003	0.004	-0.398	0	0	5	-0.009	-0.01	0.387
46	1	0	0	0	-0.003	0.004	-0.398	0	0	7.5	-0.009	-0.01	0.387
47	1	0	0	0	-0.003	0.004	-0.398	0	0	10	-0.009	-0.01	0.387
48	1	0	0	0	-0.003	0.004	-0.398	0	2.5	0	-0.009	-0.01	0.387
49	1	0	0	0	-0.003	0.004	-0.398	0	5	0	-0.009	-0.01	0.387
50	1	0	0	0	-0.003	0.004	-0.398	0	7.5	0	-0.009	-0.01	0.387
51	1	0	0	0	-0.003	0.004	-0.398	0	10	0	-0.009	-0.01	0.387
52	1	0	0	0	-0.003	0.004	-0.398	0	12.5	0	-0.009	-0.01	0.387
53	1	0	0	0	-0.003	0.004	-0.398	0	15	0	-0.009	-0.01	0.387
54	1	0	0	0	-0.003	0.004	-0.398	2	0	0	-0.009	-0.01	0.387

#	Round #	Femoral Component						Tibial Component					
		Rx	Ry	Rz	Tz	Tx	Ty	Rx	Ry	Rz	Tz	Tx	Ty
55	1	0	0	0	-0.003	0.004	-0.398	4	0	0	-0.009	-0.01	0.387
56	1	0	0	0	-0.003	0.004	-0.398	6	0	0	-0.009	-0.01	0.387
57	1	0	0	0	-0.003	0.004	-0.3955	0	0	0	-0.009	-0.01	0.387
58	1	0	0	0	-0.003	0.004	-0.393	0	0	0	-0.009	-0.01	0.387
59	1	0	0	0	-0.003	0.0065	-0.398	0	0	0	-0.009	-0.01	0.387
60	1	0	0	0	-0.003	0.009	-0.398	0	0	0	-0.009	-0.01	0.387
61	1	0	0	0	-0.0005	0.004	-0.398	0	0	0	-0.009	-0.01	0.387
62	1	0	0	0	0.002	0.004	-0.398	0	0	0	-0.009	-0.01	0.387
63	1	0	0	2.5	-0.003	0.004	-0.398	0	0	0	-0.009	-0.01	0.387
64	1	0	0	5	-0.003	0.004	-0.398	0	0	0	-0.009	-0.01	0.387
65	1	0	2.5	0	-0.003	0.004	-0.398	0	0	0	-0.009	-0.01	0.387
66	1	0	5	0	-0.003	0.004	-0.398	0	0	0	-0.009	-0.01	0.387
67	1	0	7.5	0	-0.003	0.004	-0.398	0	0	0	-0.009	-0.01	0.387
68	1	0	10	0	-0.003	0.004	-0.398	0	0	0	-0.009	-0.01	0.387
69	1	0	12.5	0	-0.003	0.004	-0.398	0	0	0	-0.009	-0.01	0.387
70	1	0	15	0	-0.003	0.004	-0.398	0	0	0	-0.009	-0.01	0.387
71	1	2	0	0	-0.003	0.004	-0.398	0	0	0	-0.009	-0.01	0.387
72	1	4	0	0	-0.003	0.004	-0.398	0	0	0	-0.009	-0.01	0.387
73	1	6	0	0	-0.003	0.004	-0.398	0	0	0	-0.009	-0.01	0.387
74	2	-6	0	0	-0.003	0.004	-0.398	-6	0	0	-0.009	-0.01	0.387
75	2	-6	0	0	-0.003	0.004	-0.398	-4	0	0	-0.009	-0.01	0.387
76	2	-6	0	0	-0.003	0.004	-0.398	-2	0	0	-0.009	-0.01	0.387
77	2	-6	0	0	-0.003	0.004	-0.398	2	0	0	-0.009	-0.01	0.387
78	2	-6	0	0	-0.003	0.004	-0.398	4	0	0	-0.009	-0.01	0.387
79	2	-6	0	0	-0.003	0.004	-0.398	6	0	0	-0.009	-0.01	0.387
80	2	-4	0	0	-0.003	0.004	-0.398	-6	0	0	-0.009	-0.01	0.387
81	2	-4	0	0	-0.003	0.004	-0.398	-4	0	0	-0.009	-0.01	0.387
82	2	-4	0	0	-0.003	0.004	-0.398	-2	0	0	-0.009	-0.01	0.387
83	2	-4	0	0	-0.003	0.004	-0.398	2	0	0	-0.009	-0.01	0.387
84	2	-4	0	0	-0.003	0.004	-0.398	4	0	0	-0.009	-0.01	0.387
85	2	-4	0	0	-0.003	0.004	-0.398	6	0	0	-0.009	-0.01	0.387
86	2	-2	0	0	-0.003	0.004	-0.398	-6	0	0	-0.009	-0.01	0.387
87	2	-2	0	0	-0.003	0.004	-0.398	-4	0	0	-0.009	-0.01	0.387
88	2	-2	0	0	-0.003	0.004	-0.398	-2	0	0	-0.009	-0.01	0.387
89	2	-2	0	0	-0.003	0.004	-0.398	2	0	0	-0.009	-0.01	0.387
90	2	-2	0	0	-0.003	0.004	-0.398	4	0	0	-0.009	-0.01	0.387
91	2	-2	0	0	-0.003	0.004	-0.398	6	0	0	-0.009	-0.01	0.387
92	2	0	-20	0	-0.003	0.004	-0.398	0	-20	0	-0.009	-0.01	0.387
93	2	0	-20	0	-0.003	0.004	-0.398	0	-15	0	-0.009	-0.01	0.387

#	Round #	Femoral Component						Tibial Component					
		Rx	Ry	Rz	Tz	Tx	Ty	Rx	Ry	Rz	Tz	Tx	Ty
94	2	0	-20	0	-0.003	0.004	-0.398	0	-10	0	-0.009	-0.01	0.387
95	2	0	-20	0	-0.003	0.004	-0.398	0	-5	0	-0.009	-0.01	0.387
96	2	0	-20	0	-0.003	0.004	-0.398	0	5	0	-0.009	-0.01	0.387
97	2	0	-20	0	-0.003	0.004	-0.398	0	10	0	-0.009	-0.01	0.387
98	2	0	-20	0	-0.003	0.004	-0.398	0	15	0	-0.009	-0.01	0.387
99	2	0	-20	0	-0.003	0.004	-0.398	0	20	0	-0.009	-0.01	0.387
100	2	0	-15	0	-0.003	0.004	-0.398	0	-20	0	-0.009	-0.01	0.387
101	2	0	-15	0	-0.003	0.004	-0.398	0	-15	0	-0.009	-0.01	0.387
102	2	0	-15	0	-0.003	0.004	-0.398	0	-10	0	-0.009	-0.01	0.387
103	2	0	-15	0	-0.003	0.004	-0.398	0	-5	0	-0.009	-0.01	0.387
104	2	0	-15	0	-0.003	0.004	-0.398	0	5	0	-0.009	-0.01	0.387
105	2	0	-15	0	-0.003	0.004	-0.398	0	10	0	-0.009	-0.01	0.387
106	2	0	-15	0	-0.003	0.004	-0.398	0	15	0	-0.009	-0.01	0.387
107	2	0	-15	0	-0.003	0.004	-0.398	0	20	0	-0.009	-0.01	0.387
108	2	0	-10	0	-0.003	0.004	-0.398	0	-20	0	-0.009	-0.01	0.387
109	2	0	-10	0	-0.003	0.004	-0.398	0	-15	0	-0.009	-0.01	0.387
110	2	0	-10	0	-0.003	0.004	-0.398	0	-10	0	-0.009	-0.01	0.387
111	2	0	-10	0	-0.003	0.004	-0.398	0	-5	0	-0.009	-0.01	0.387
112	2	0	-10	0	-0.003	0.004	-0.398	0	5	0	-0.009	-0.01	0.387
113	2	0	-10	0	-0.003	0.004	-0.398	0	10	0	-0.009	-0.01	0.387
114	2	0	-10	0	-0.003	0.004	-0.398	0	15	0	-0.009	-0.01	0.387
115	2	0	-10	0	-0.003	0.004	-0.398	0	20	0	-0.009	-0.01	0.387
116	2	0	-5	0	-0.003	0.004	-0.398	0	-20	0	-0.009	-0.01	0.387
117	2	0	-5	0	-0.003	0.004	-0.398	0	-15	0	-0.009	-0.01	0.387
118	2	0	-5	0	-0.003	0.004	-0.398	0	-10	0	-0.009	-0.01	0.387
119	2	0	-5	0	-0.003	0.004	-0.398	0	-5	0	-0.009	-0.01	0.387
120	2	0	-5	0	-0.003	0.004	-0.398	0	5	0	-0.009	-0.01	0.387
121	2	0	-5	0	-0.003	0.004	-0.398	0	10	0	-0.009	-0.01	0.387
122	2	0	-5	0	-0.003	0.004	-0.398	0	15	0	-0.009	-0.01	0.387
123	2	0	-5	0	-0.003	0.004	-0.398	0	20	0	-0.009	-0.01	0.387
124	2	0	0	-10	-0.003	0.004	-0.398	0	0	-5	-0.009	-0.01	0.387
125	2	0	0	-10	-0.003	0.004	-0.398	0	0	-2.5	-0.009	-0.01	0.387
126	2	0	0	-10	-0.003	0.004	-0.398	0	0	2.5	-0.009	-0.01	0.387
127	2	0	0	-10	-0.003	0.004	-0.398	0	0	5	-0.009	-0.01	0.387
128	2	0	0	-10	-0.003	0.004	-0.398	0	0	7.5	-0.009	-0.01	0.387
129	2	0	0	-10	-0.003	0.004	-0.398	0	0	10	-0.009	-0.01	0.387
130	2	0	0	-7.5	-0.003	0.004	-0.398	0	0	-5	-0.009	-0.01	0.387
131	2	0	0	-7.5	-0.003	0.004	-0.398	0	0	-2.5	-0.009	-0.01	0.387
132	2	0	0	-7.5	-0.003	0.004	-0.398	0	0	2.5	-0.009	-0.01	0.387

#	Round #	Femoral Component						Tibial Component					
		Rx	Ry	Rz	Tz	Tx	Ty	Rx	Ry	Rz	Tz	Tx	Ty
133	2	0	0	-7.5	-0.003	0.004	-0.398	0	0	5	-0.009	-0.01	0.387
134	2	0	0	-7.5	-0.003	0.004	-0.398	0	0	7.5	-0.009	-0.01	0.387
135	2	0	0	-7.5	-0.003	0.004	-0.398	0	0	10	-0.009	-0.01	0.387
136	2	0	0	-5	-0.003	0.004	-0.398	0	0	-5	-0.009	-0.01	0.387
137	2	0	0	-5	-0.003	0.004	-0.398	0	0	-2.5	-0.009	-0.01	0.387
138	2	0	0	-5	-0.003	0.004	-0.398	0	0	2.5	-0.009	-0.01	0.387
139	2	0	0	-5	-0.003	0.004	-0.398	0	0	5	-0.009	-0.01	0.387
140	2	0	0	-5	-0.003	0.004	-0.398	0	0	7.5	-0.009	-0.01	0.387
141	2	0	0	-5	-0.003	0.004	-0.398	0	0	10	-0.009	-0.01	0.387
142	2	0	0	-2.5	-0.003	0.004	-0.398	0	0	-5	-0.009	-0.01	0.387
143	2	0	0	-2.5	-0.003	0.004	-0.398	0	0	-2.5	-0.009	-0.01	0.387
144	2	0	0	-2.5	-0.003	0.004	-0.398	0	0	2.5	-0.009	-0.01	0.387
145	2	0	0	-2.5	-0.003	0.004	-0.398	0	0	5	-0.009	-0.01	0.387
146	2	0	0	-2.5	-0.003	0.004	-0.398	0	0	7.5	-0.009	-0.01	0.387
147	2	0	0	-2.5	-0.003	0.004	-0.398	0	0	10	-0.009	-0.01	0.387
148	2	0	0	2.5	-0.003	0.004	-0.398	0	0	-5	-0.009	-0.01	0.387
149	2	0	0	2.5	-0.003	0.004	-0.398	0	0	-2.5	-0.009	-0.01	0.387
150	2	0	0	2.5	-0.003	0.004	-0.398	0	0	2.5	-0.009	-0.01	0.387
151	2	0	0	2.5	-0.003	0.004	-0.398	0	0	5	-0.009	-0.01	0.387
152	2	0	0	2.5	-0.003	0.004	-0.398	0	0	7.5	-0.009	-0.01	0.387
153	2	0	0	2.5	-0.003	0.004	-0.398	0	0	10	-0.009	-0.01	0.387
154	2	0	0	5	-0.003	0.004	-0.398	0	0	-5	-0.009	-0.01	0.387
155	2	0	0	5	-0.003	0.004	-0.398	0	0	-2.5	-0.009	-0.01	0.387
156	2	0	0	5	-0.003	0.004	-0.398	0	0	2.5	-0.009	-0.01	0.387
157	2	0	0	5	-0.003	0.004	-0.398	0	0	5	-0.009	-0.01	0.387
158	2	0	0	5	-0.003	0.004	-0.398	0	0	7.5	-0.009	-0.01	0.387
159	2	0	0	5	-0.003	0.004	-0.398	0	0	10	-0.009	-0.01	0.387
160	2	0	5	0	-0.003	0.004	-0.398	0	-20	0	-0.009	-0.01	0.387
161	2	0	5	0	-0.003	0.004	-0.398	0	-15	0	-0.009	-0.01	0.387
162	2	0	5	0	-0.003	0.004	-0.398	0	-10	0	-0.009	-0.01	0.387
163	2	0	5	0	-0.003	0.004	-0.398	0	-5	0	-0.009	-0.01	0.387
164	2	0	5	0	-0.003	0.004	-0.398	0	5	0	-0.009	-0.01	0.387
165	2	0	5	0	-0.003	0.004	-0.398	0	10	0	-0.009	-0.01	0.387
166	2	0	5	0	-0.003	0.004	-0.398	0	15	0	-0.009	-0.01	0.387
167	2	0	5	0	-0.003	0.004	-0.398	0	20	0	-0.009	-0.01	0.387
168	2	0	10	0	-0.003	0.004	-0.398	0	-20	0	-0.009	-0.01	0.387
169	2	0	10	0	-0.003	0.004	-0.398	0	-15	0	-0.009	-0.01	0.387
170	2	0	10	0	-0.003	0.004	-0.398	0	-10	0	-0.009	-0.01	0.387
171	2	0	10	0	-0.003	0.004	-0.398	0	-5	0	-0.009	-0.01	0.387

#	Round #	Femoral Component						Tibial Component					
		Rx	Ry	Rz	Tz	Tx	Ty	Rx	Ry	Rz	Tz	Tx	Ty
172	2	0	10	0	-0.003	0.004	-0.398	0	5	0	-0.009	-0.01	0.387
173	2	0	10	0	-0.003	0.004	-0.398	0	10	0	-0.009	-0.01	0.387
174	2	0	10	0	-0.003	0.004	-0.398	0	15	0	-0.009	-0.01	0.387
175	2	0	10	0	-0.003	0.004	-0.398	0	20	0	-0.009	-0.01	0.387
176	2	0	15	0	-0.003	0.004	-0.398	0	-20	0	-0.009	-0.01	0.387
177	2	0	15	0	-0.003	0.004	-0.398	0	-15	0	-0.009	-0.01	0.387
178	2	0	15	0	-0.003	0.004	-0.398	0	-10	0	-0.009	-0.01	0.387
179	2	0	15	0	-0.003	0.004	-0.398	0	-5	0	-0.009	-0.01	0.387
180	2	0	15	0	-0.003	0.004	-0.398	0	5	0	-0.009	-0.01	0.387
181	2	0	15	0	-0.003	0.004	-0.398	0	10	0	-0.009	-0.01	0.387
182	2	0	15	0	-0.003	0.004	-0.398	0	15	0	-0.009	-0.01	0.387
183	2	0	15	0	-0.003	0.004	-0.398	0	20	0	-0.009	-0.01	0.387
184	2	0	20	0	-0.003	0.004	-0.398	0	-20	0	-0.009	-0.01	0.387
185	2	0	20	0	-0.003	0.004	-0.398	0	-15	0	-0.009	-0.01	0.387
186	2	0	20	0	-0.003	0.004	-0.398	0	-10	0	-0.009	-0.01	0.387
187	2	0	20	0	-0.003	0.004	-0.398	0	-5	0	-0.009	-0.01	0.387
188	2	0	20	0	-0.003	0.004	-0.398	0	5	0	-0.009	-0.01	0.387
189	2	0	20	0	-0.003	0.004	-0.398	0	10	0	-0.009	-0.01	0.387
190	2	0	20	0	-0.003	0.004	-0.398	0	15	0	-0.009	-0.01	0.387
191	2	0	20	0	-0.003	0.004	-0.398	0	20	0	-0.009	-0.01	0.387
192	2	2	0	0	-0.003	0.004	-0.398	-6	0	0	-0.009	-0.01	0.387
193	2	2	0	0	-0.003	0.004	-0.398	-4	0	0	-0.009	-0.01	0.387
194	2	2	0	0	-0.003	0.004	-0.398	-2	0	0	-0.009	-0.01	0.387
195	2	2	0	0	-0.003	0.004	-0.398	2	0	0	-0.009	-0.01	0.387
196	2	2	0	0	-0.003	0.004	-0.398	4	0	0	-0.009	-0.01	0.387
197	2	2	0	0	-0.003	0.004	-0.398	6	0	0	-0.009	-0.01	0.387
198	2	4	0	0	-0.003	0.004	-0.398	-6	0	0	-0.009	-0.01	0.387
199	2	4	0	0	-0.003	0.004	-0.398	-4	0	0	-0.009	-0.01	0.387
200	2	4	0	0	-0.003	0.004	-0.398	-2	0	0	-0.009	-0.01	0.387
201	2	4	0	0	-0.003	0.004	-0.398	2	0	0	-0.009	-0.01	0.387
202	2	4	0	0	-0.003	0.004	-0.398	4	0	0	-0.009	-0.01	0.387
203	2	4	0	0	-0.003	0.004	-0.398	6	0	0	-0.009	-0.01	0.387
204	2	6	0	0	-0.003	0.004	-0.398	-6	0	0	-0.009	-0.01	0.387
205	2	6	0	0	-0.003	0.004	-0.398	-4	0	0	-0.009	-0.01	0.387
206	2	6	0	0	-0.003	0.004	-0.398	-2	0	0	-0.009	-0.01	0.387
207	2	6	0	0	-0.003	0.004	-0.398	2	0	0	-0.009	-0.01	0.387
208	2	6	0	0	-0.003	0.004	-0.398	4	0	0	-0.009	-0.01	0.387
209	2	6	0	0	-0.003	0.004	-0.398	6	0	0	-0.009	-0.01	0.387

Appendix B

Coefficients to describe estimated kinematics curve

Table 14. Coefficients to describe knee kinematics curve.

	Knee V/V_A	Knee V/V_B	Knee V/V_C
Constant	-4.597E-05	1.163E-02	-4.605E-01
fc_V/V	-4.949E-05	-8.528E-03	1.194E+00
fc_I/E	1.343E-04	-2.399E-02	8.947E-02
fc_F/R	5.726E-07	-4.496E-05	1.965E-03
tc_V/V	-3.810E-05	3.614E-03	-1.048E+00
tc_I/E	1.390E-05	-1.457E-03	5.443E-02
tc_A/P	-1.443E-05	1.148E-03	-1.756E-02
fc_V/V*fc_V/V	1.439E-06	-2.913E-04	1.247E-02
fc_I/E*fc_I/E	6.932E-08	4.931E-06	-1.327E-04
fc_F/R*fc_F/R	-4.900E-07	6.374E-05	-1.650E-03
tc_V/V*tc_V/V	2.933E-06	-3.742E-04	1.020E-02
tc_I/E*tc_I/E	5.514E-07	-6.233E-05	1.511E-03
tc_A/P*tc_A/P	2.109E-06	-2.393E-04	6.110E-03

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